WEST

Freeform Search

Database: US Patents Full-Text Database
US Pre-Grant Publication Full-Text Database
JPO Abstracts Database
EPO Abstracts Database
Derwent World Patents Index
IBM Technical Disclosure Bulletins

Term: L18 and (zero)

Display: 100 Documents in Display Format: - Starting with Number 1

Generate: ○ Hit List ◉ Hit Count ○ Side by Side ○ Image

Search | Clear | Help | Logout | Interrupt

Main Menu | Show S Numbers | Edit S Numbers | Preferences | Cases

Search History

DATE: Wednesday, September 04, 2002 Printable Copy Create Case

Set Name side by side	Query	Hit Count	Set Name result set
	DB=USPT,PGPB,JPAB,EPAB,DWPI,TDBD; PLUR=YES; OP=ADJ		
L19	L18 and (zero)	3	L19
L18	L17 and (net)	6	L18
L17	L16 and (mutual$3)	9	L17
L16	L15 and (induct$6)	12	L16
L15	L14 and (parallel or perpendicular$2 or orthogonal$2)	12	L15
L14	L13 and (mode)	12	L14
L13	L12 and ((zero or "0" or "no" or null or null$6 or cancel$5 or eliminat$6) with coupl$5)	17	L13
L12	L11 and (coupl$5)	48	L12
L11	L10 and (current with (path or pattern or winding))	76	L11
L10	L9 and (current)	122	L10
L9	L8 and ((plurality or group or set or array) with coil)	126	L9
L8	L6 and ((flux or zero-flux or "zero flux" or "0-flux" or "0 flux" or ("no" with flux)) with (contour$4 or profile or trajector$4 or path))	172	L8
L7	L2 and ((flux or zero-flux or "zero flux" or "0-flux" or "0 flux" or ("no" with flux)) with (contour$4 or profile or trajector$4 or path))	353	L7
L6	L5 and (contour$4 or profile or trajector$4 or path)	389	L6
L5	L4 and ((pair or two) with coils)	594	L5
L4	L3 and ((single or one or individual) with coil)	735	L4
L3	L2 and (flux or zero-flux or "zero flux" or "0-flux" or "0 flux" or ("no" with flux))	2602	L3
L2	L1 and (opposite or revers$4 or return$4 or clockwise or "counter clockwise" or counter-clockwise)	42934	L2
L1	((magnetic adj resonance) or MRI or NMR)	132190	L1

END OF SEARCH HISTORY

WEST

Generate Collection | Print

Search Results - Record(s) 1 through 12 of 12 returned.

☐ 1. Document ID: US 6426917 B1

L16: Entry 1 of 12 File: USPT Jul 30, 2002

US-PAT-NO: 6426917
DOCUMENT-IDENTIFIER: US 6426917 B1

TITLE: Reservoir monitoring through modified casing joint

DATE-ISSUED: July 30, 2002

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Tabanou; Jacques	Houston	TX		
Ciglenec; Reinhart	Houston	TX		
Eckersley; Clive	Cairo			EG
Chouzenoux; Christian	St. Cloud			FR

US-CL-CURRENT: 367/82; 324/333, 324/338, 340/854.4, 340/854.6

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 2. Document ID: US 6285189 B1

L16: Entry 2 of 12 File: USPT Sep 4, 2001

US-PAT-NO: 6285189
DOCUMENT-IDENTIFIER: US 6285189 B1

TITLE: Millipede coils

DATE-ISSUED: September 4, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wong; Wai Ha	San Jose	CA		

US-CL-CURRENT: 324/318; 324/321, 335/299, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 3. Document ID: US 6060882 A

L16: Entry 3 of 12 File: USPT May 9, 2000

US-PAT-NO: 6060882
DOCUMENT-IDENTIFIER: US 6060882 A

TITLE: Low-inductance transverse litz foil coils

DATE-ISSUED: May 9, 2000

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Doty; F. David	Columbia	SC		

US-CL-CURRENT: 324/318; 324/319, 324/322, 600/421

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

---

4. Document ID: US 5886596 A

L16: Entry 4 of 12 File: USPT Mar 23, 1999

US-PAT-NO: 5886596
DOCUMENT-IDENTIFIER: US 5886596 A

TITLE: Radio frequency volume coils for imaging and spectroscopy

DATE-ISSUED: March 23, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Vaughan, Jr.; John Thomas	Lynnfield	MA		

US-CL-CURRENT: 333/219; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

---

5. Document ID: US 5798679 A

L16: Entry 5 of 12 File: USPT Aug 25, 1998

US-PAT-NO: 5798679
DOCUMENT-IDENTIFIER: US 5798679 A

TITLE: Magnetic flux bending devices

DATE-ISSUED: August 25, 1998

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Pissanetzky; Sergio	The Woodlands	TX		

US-CL-CURRENT: 335/299

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

---

6. Document ID: US 5557247 A

L16: Entry 6 of 12 File: USPT Sep 17, 1996

US-PAT-NO: 5557247
DOCUMENT-IDENTIFIER: US 5557247 A

TITLE: Radio frequency volume coils for imaging and spectroscopy

DATE-ISSUED: September 17, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Vaughn, Jr.; John T.	Birmingham	AL		

US-CL-CURRENT: 333/219; 324/318, 324/322, 333/222, 333/227

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

---

7. Document ID: US 5521506 A

L16: Entry 7 of 12 File: USPT May 28, 1996

US-PAT-NO: 5521506
DOCUMENT-IDENTIFIER: US 5521506 A

TITLE: Orthogonal adjustment of magnetic resonance surface coils

DATE-ISSUED: May 28, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George	Novelty	OH		
Reid; Eric	Monroeville	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

---

8. Document ID: US 5477146 A

L16: Entry 8 of 12 File: USPT Dec 19, 1995

US-PAT-NO: 5477146
DOCUMENT-IDENTIFIER: US 5477146 A

TITLE: NMR adjustable volume array

DATE-ISSUED: December 19, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jones; Randall W.	Elkhorn	NE		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 9. Document ID: US 5202635 A

L16: Entry 9 of 12 File: USPT Apr 13, 1993

US-PAT-NO: 5202635
DOCUMENT-IDENTIFIER: US 5202635 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: April 13, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Philadelphia	PA		
Murphy-Boesch; Joseph	Lafayette Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 10. Document ID: US 5194811 A

L16: Entry 10 of 12 File: USPT Mar 16, 1993

US-PAT-NO: 5194811
DOCUMENT-IDENTIFIER: US 5194811 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: March 16, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hill	PA		
Srinivasan; Ravi	Philadelphia	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 11. Document ID: US 4799016 A

L16: Entry 11 of 12 File: USPT Jan 17, 1989

US-PAT-NO: 4799016
DOCUMENT-IDENTIFIER: US 4799016 A

TITLE: Dual frequency NMR surface coil

DATE-ISSUED: January 17, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Rezvani; Behrooz	Shorewood	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 12. Document ID: US 4694255 A

L16: Entry 12 of 12 File: USPT Sep 15, 1987

US-PAT-NO: 4694255
DOCUMENT-IDENTIFIER: US 4694255 A

TITLE: Radio frequency field coil for NMR

DATE-ISSUED: September 15, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hayes; Cecil E.	Wauwatosa	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

Generate Collection | Print

Term	Documents
INDUCT$6	0
INDUCT.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1121
INDUCTA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	4
INDUCTABLE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	7
INDUCTABLY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	9
INDUCTACE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	7
INDUCTACES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2
INDUCTACNCE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
INDUCTALLOY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	3
INDUCTAMCES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
INDUCTAME.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
(L15 AND (INDUCT$6)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	12

There are more results than shown above. Click here to view the entire set.

Display Format: Change Format

Previous Page Next Page

WEST

Generate Collection | Print

Search Results - Record(s) 1 through 9 of 9 returned.

---

1. Document ID: US 6285189 B1

L17: Entry 1 of 9 File: USPT Sep 4, 2001

US-PAT-NO: 6285189
DOCUMENT-IDENTIFIER: US 6285189 B1

TITLE: Millipede coils

DATE-ISSUED: September 4, 2001

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Wong; Wai Ha	San Jose	CA		

US-CL-CURRENT: 324/318; 324/321, 335/299, 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw Desc | Image

---

2. Document ID: US 5886596 A

L17: Entry 2 of 9 File: USPT Mar 23, 1999

US-PAT-NO: 5886596
DOCUMENT-IDENTIFIER: US 5886596 A

TITLE: Radio frequency volume coils for imaging and spectroscopy

DATE-ISSUED: March 23, 1999

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Vaughan, Jr.; John Thomas	Lynnfield	MA		

US-CL-CURRENT: 333/219; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | Claims | KWIC | Draw Desc | Image

---

3. Document ID: US 5557247 A

L17: Entry 3 of 9 File: USPT Sep 17, 1996

US-PAT-NO: 5557247
DOCUMENT-IDENTIFIER: US 5557247 A

TITLE: Radio frequency volume coils for imaging and spectroscopy

DATE-ISSUED: September 17, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Vaughn, Jr.; John T.	Birmingham	AL		

US-CL-CURRENT: 333/219; 324/318, 324/322, 333/222, 333/227

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 4. Document ID: US 5521506 A

L17: Entry 4 of 9 File: USPT May 28, 1996

US-PAT-NO: 5521506
DOCUMENT-IDENTIFIER: US 5521506 A

TITLE: Orthogonal adjustment of magnetic resonance surface coils

DATE-ISSUED: May 28, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George	Novelty	OH		
Reid; Eric	Monroeville	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 5. Document ID: US 5477146 A

L17: Entry 5 of 9 File: USPT Dec 19, 1995

US-PAT-NO: 5477146
DOCUMENT-IDENTIFIER: US 5477146 A

TITLE: NMR adjustable volume array

DATE-ISSUED: December 19, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jones; Randall W.	Elkhorn	NE		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

6. Document ID: US 5202635 A

L17: Entry 6 of 9 File: USPT Apr 13, 1993

US-PAT-NO: 5202635
DOCUMENT-IDENTIFIER: US 5202635 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: April 13, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Philadelphia	PA		
Murphy-Boesch; Joseph	Lafayette Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

7. Document ID: US 5194811 A

L17: Entry 7 of 9 File: USPT Mar 16, 1993

US-PAT-NO: 5194811
DOCUMENT-IDENTIFIER: US 5194811 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: March 16, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hill	PA		
Srinivasan; Ravi	Philadelphia	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

8. Document ID: US 4799016 A

L17: Entry 8 of 9 File: USPT Jan 17, 1989

US-PAT-NO: 4799016
DOCUMENT-IDENTIFIER: US 4799016 A

TITLE: Dual frequency NMR surface coil

DATE-ISSUED: January 17, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Rezvani; Behrooz	Shorewood	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

---

9. Document ID: US 4694255 A

L17: Entry 9 of 9 File: USPT Sep 15, 1987

US-PAT-NO: 4694255
DOCUMENT-IDENTIFIER: US 4694255 A

TITLE: Radio frequency field coil for NMR

DATE-ISSUED: September 15, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hayes; Cecil E.	Wauwatosa	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

Generate Collection | Print

Term	Documents
MUTUAL$3	0
MUTUAL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	171410
MUTUALA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
MUTUALISM.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	18
MUTUALITE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2
MUTUALITY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	86
MUTUALIY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
MUTUALIZE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	2
MUTUALL.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	24
MUTUALLA.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
MUTUALLAY.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	1
(L16 AND (MUTUAL$3)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	9

There are more results than shown above. Click here to view the entire set.

Display Format: - Change Format

Previous Page   Next Page

WEST

Generate Collection | Print

Search Results - Record(s) 1 through 6 of 6 returned.

☐ 1. Document ID: US 5521506 A

L18: Entry 1 of 6 File: USPT May 28, 1996

US-PAT-NO: 5521506
DOCUMENT-IDENTIFIER: US 5521506 A

TITLE: Orthogonal adjustment of magnetic resonance surface coils

DATE-ISSUED: May 28, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George	Novelty	OH		
Reid; Eric	Monroeville	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 2. Document ID: US 5477146 A

L18: Entry 2 of 6 File: USPT Dec 19, 1995

US-PAT-NO: 5477146
DOCUMENT-IDENTIFIER: US 5477146 A

TITLE: NMR adjustable volume array

DATE-ISSUED: December 19, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jones; Randall W.	Elkhorn	NE		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 3. Document ID: US 5202635 A

L18: Entry 3 of 6 File: USPT Apr 13, 1993

US-PAT-NO: 5202635

DOCUMENT-IDENTIFIER: US 5202635 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: April 13, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Srinivasan; Ravi	Philadelphia	PA		
Murphy-Boesch; Joseph	Lafayette Hills	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 4. Document ID: US 5194811 A

L18: Entry 4 of 6 File: USPT Mar 16, 1993

US-PAT-NO: 5194811
DOCUMENT-IDENTIFIER: US 5194811 A

TITLE: Radio frequency volume resonator for nuclear magnetic resonance

DATE-ISSUED: March 16, 1993

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Murphy-Boesch; Joseph	Lafayette Hill	PA		
Srinivasan; Ravi	Philadelphia	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 5. Document ID: US 4799016 A

L18: Entry 5 of 6 File: USPT Jan 17, 1989

US-PAT-NO: 4799016
DOCUMENT-IDENTIFIER: US 4799016 A

TITLE: Dual frequency NMR surface coil

DATE-ISSUED: January 17, 1989

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Rezvani; Behrooz	Shorewood	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

6. Document ID: US 4694255 A

L18: Entry 6 of 6 File: USPT Sep 15, 1987

US-PAT-NO: 4694255
DOCUMENT-IDENTIFIER: US 4694255 A

TITLE: Radio frequency field coil for NMR

DATE-ISSUED: September 15, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hayes; Cecil E.	Wauwatosa	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

Generate Collection | Print

Term	Documents
NET.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	192853
NETS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	17011
(17 AND NET).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	6
(L17 AND (NET)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	6

Display Format: - Change Format

Previous Page Next Page

WEST

Generate Collection | Print

Search Results - Record(s) 1 through 3 of 3 returned.

☐ 1. Document ID: US 5521506 A

L19: Entry 1 of 3 File: USPT May 28, 1996

US-PAT-NO: 5521506
DOCUMENT-IDENTIFIER: US 5521506 A

TITLE: Orthogonal adjustment of magnetic resonance surface coils

DATE-ISSUED: May 28, 1996

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Misic; George	Novelty	OH		
Reid; Eric	Monroeville	PA		

US-CL-CURRENT: 324/322; 324/318

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 2. Document ID: US 5477146 A

L19: Entry 2 of 3 File: USPT Dec 19, 1995

US-PAT-NO: 5477146
DOCUMENT-IDENTIFIER: US 5477146 A

TITLE: NMR adjustable volume array

DATE-ISSUED: December 19, 1995

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Jones; Randall W.	Elkhorn	NE		

US-CL-CURRENT: 324/318; 600/422

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

☐ 3. Document ID: US 4694255 A

L19: Entry 3 of 3 File: USPT Sep 15, 1987

US-PAT-NO: 4694255

DOCUMENT-IDENTIFIER: US 4694255 A

TITLE: Radio frequency field coil for NMR

DATE-ISSUED: September 15, 1987

INVENTOR-INFORMATION:

NAME	CITY	STATE	ZIP CODE	COUNTRY
Hayes; Cecil E.	Wauwatosa	WI		

US-CL-CURRENT: 324/318; 324/322

Full | Title | Citation | Front | Review | Classification | Date | Reference | Sequences | Attachments | KWIC
Draw Desc | Image

Generate Collection | Print

Term	Documents
ZERO.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	643297
ZEROES.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	9270
ZEROS.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	25520
ZEROE.DWPI,TDBD,EPAB,JPAB,USPT,PGPB.	8
(18 AND ZERO).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	3
(L18 AND (ZERO)).USPT,PGPB,JPAB,EPAB,DWPI,TDBD.	3

Display Format: - Change Format

Previous Page Next Page

L19: Entry 1 of 3 File: USPT May 28, 1996

DOCUMENT-IDENTIFIER: US 5521506 A
TITLE: Orthogonal adjustment of magnetic resonance surface coils

Abstract Text (1):
An NMR magnetic coil system (10) is disclosed wherein the isolation between the coils (16, 18) can be adjusted to decrease or virtually eliminate the coupling between quadrature magnetic resonance imaging coils (16, 18). At least one of the coils (16) is separated into parallel segments (20, 22), located in a critically overlapped area. The capacitance of the segments is adjusted by a differential capacitor (76) to vary the ratio of the RF current flowing through the parallel segments. Appropriate adjustment of the capacitance of these paths (20, 22) causes a sharing of the appropriate amount of the out of phase flux to cancel the balance of the shared flux and therefore results in a net mutual inductance of zero.

Brief Summary Text (2):
This invention relates generally to the field of magnetic resonance imaging technology. Specifically this invention relates to several improved methods for the electrical adjustment of two or more magnetic resonance coils to assure the proper isolation and/or orthogonal relationship of the coil fields in order to increase the signal to noise ratio of the magnetic resonance signal.

Brief Summary Text (4):
Quadrature magnetic resonance imaging coils, and more recently, multicoil systems using a plurality of independent data acquisition channels, are generally known in the art. Quadrature magnetic resonance systems offer advantages over previous magnetic resonance imaging techniques in that they provide a better signal to noise ratio by utilizing both component vectors of the circularly polarized field of the magnetic resonance phenomenon, and lower RF transmitter power requirements when used as transmit coils. Multicoil systems offer some or all of the above-noted advantages, plus additional advantages in enhancing the imaging signal to noise ratio due to the reduced imaging volume of each independent coil and data acquisition path in the multicoil system. However, when these systems are used for magnetic resonance imaging, the isolation of the signal currents in one coil mode or coil system from currents in the other mode or coil system must be at a high level to obtain the benefits of quadrature operation, or multicoil operation.

Brief Summary Text (5):
Those skilled in the art will appreciate that it is desirable to reduce or eliminate the inductive coupling between the two coil systems forming the RF quadrature coil used in a magnetic resonance imaging system in order to solve these and other problems. Additionally, it is desirable to reduce or eliminate the inductive coupling betwixt the various coil systems in a multicoil configuration. Ideally there should be no inductive coupling between the coil systems comprising the RF quadrature coil or multicoil system. Previously the adjustment of such coils to minimize the coupling between the coils was accomplished by either the physical movement of the coils or the physical adjustment of a variable element to electrically accomplish the same result.

Brief Summary Text (6):
Changing a single element generally alters the tuning or other coil parameters. In the past, adjusting the isolation or orthogonality of a coil has yielded undesirable secondary adjustment of one or more other coil parameters. Further, if physical adjustment of the location of the coils is employed to accomplish this result, many coil formations are eliminated as a practical matter, thereby dramatically decreasing the versatility of these systems.

Brief Summary Text (7):
While the conventional devices have made significant advances in the art of magnetic resonance imaging, it is clear that much more versatile and useful magnetic resonance imaging systems will result from a quadrature magnetic resonance coil system that can be adjusted, or remotely adjusted, to only optimize the isolation of the coil elements without affecting other operational characteristics.

Brief Summary Text (9):
The above-noted quadrature coil adjustment problems of conventional systems are solved by the present invention. In accordance with one aspect of the invention, a magnetic resonance coil system includes a first coil, and a second coil having first and second segments configured in parallel. The second coil physically overlaps the first coil. A differential capacitor is provided which contributes a first capacitance in series with the first parallel segment of the second coil and a second capacitance in series with the second parallel segment of the second coil. The differential capacitor is operable to vary the first and second capacitances so as to vary the ratio of RF current present in the two segments.

Brief Summary Text (10):
Preferably, adjusting the differential capacitor to increase the first capacitance will proportionally reduce the second capacitance, and vice versa. This variation in the first and second capacitances changes the relative current flow in the first and second sections of the split conductor or parallel segment, and thereby adjusts the orthogonality of the second coil field with respect to the first by altering the proportion of overlapped to non-overlapped areas. The total net capacitance remains essentially constant, and thus the tuning and matching characteristics of the coil system are not materially affected.

Brief Summary Text (11):
In a first preferred embodiment, the capacitors are remotely operable to vary the first and second capacitances such that the capacitors and associated circuitry do not have to be physically disturbed. With the use of a differential capacitor, the ratio of capacitance may further be varied such when the first capacitance is increased, the second capacitance is reduced. Adjusting the capacitance in this manner changes the ratio of the RF current flowing through each segment, causing a sharing of the proper amount of out-of-phase flux to cancel the balance of the shared in-phase flux between the first and second coils. When the differential capacitor is set to an appropriate position, the net shared flux, and therefore the mutual inductance, approach zero.

Brief Summary Text (12):
In another preferred embodiment, the differential capacitor comprises first and second varactor diodes which are disposed in respective parallel segments of the second coil. Variable voltage circuitry is coupled to PN junctions of these diodes to vary the reverse bias on the varactor diodes which thereby adjusts the first and second capacitances. A plurality of potentiometers or other voltage adjustment means within the variable voltage circuitry are supplied in order to vary the voltages appearing at these diodes.

Brief Summary Text (13):
The present invention confers a principal technical advantage in that the orthogonality of RF quadrature coils can be precisely adjusted without disturbing the coils either by making physical adjustments to variable capacitors or by physically moving the coils themselves. Differential adjustment of the voltages on the varactor diodes provides orthogonality adjustment, whilst common mode adjustment of the voltages can be employed to provide tuning adjustment. The present invention has applications in many different types of quadrature magnetic resonance surface coils and/or multicoil systems.

Drawing Description Text (5):
FIG. 3 is an electrical schematic diagram illustrating another NMR quadrature coil pair, varactor diodes making up a differential capacitor for adjusting coil orthogonality, and bias circuitry for control of the varactor diodes;

Detailed Description Text (2):
In accordance with the present invention a quadrature magnetic resonance surface coil assembly is disclosed which includes first and second coils with generally perpendicular fields. The first and second coils are mounted parallel to each other

and in a preferred embodiment share a critically overlapped area. In the preferred embodiment, each of the coils is intentionally split into two separate parallel segments within the critically overlapped area. It is apparent, however, that the invention would provide satisfactory results if only one of the coils was divided into separate parallel segments,

Detailed Description Text (3):
The coils are shaped and positioned so that the mutual inductance and therefore the majority of the coupling between the loops are minimized. This is usually accomplished by physically overlapping a critical portion of the area enclosed by the loops, such that the vector sum of all flux acting on one coil due to the other coil is near zero. However, the positioning of the loops is highly critical and is too sensitive to allow mass-produced devices to perform well without further individual adjustment to compensate for tolerances in physical dimensions and in values of electrical components,

Detailed Description Text (4):
The net mutual inductance of coils in a particular individual unit can be reduced to a level approaching zero when a differential capacitor or reactor is placed in series with the parallel segments and is appropriately adjusted. The results of such an individual adjustment can be measured using a network analyzer set to display the S21 (transmission) parameter. With one coil on each analyzer port, the coils are adjusted to minimize the value of S21. In the preferred embodiments, the differential element is a capacitor that can be adjusted to vary the ratio of current in each of the parallel segments of the coils. Isolation and orthogonality between the coils can be adjusted if the adjustable element performs the function of a differential capacitor.

Detailed Description Text (5):
Adjusting the capacitance in series with each of the parallel segments varies the ratio of the RF current flowing through each segment of the loop. This accomplishes an effect equivalent to physically moving the loops to change the amount of overlap area and shared flux. The resulting effective change in the overlapped area effectively changes the ratio of the out-of-phase flux to in-phase flux of the loops.

Detailed Description Text (6):
In another embodiment, the first and second coils are disposed such that a principal electromagnetic field of the first coil is orthogonal to a principal electromagnetic field of the second coil. The split conductor segments and associated differential reactive element serve to adjust the relative orthogonality of the fields of one or more of the coils with respect to the field from another coil or coils. In yet a further embodiment, the split parallel conductors may lay either wholly or partially outside of a critical overlap area. The coil systems according to the invention may be configured as multicoil or phased array coil systems. Some or all of the component coil subsystems may be quadrature coils. Below are described several examples of how the coils may be configured.

Detailed Description Text (7):
FIG. 1 is a schematic diagram illustrating a representative NMR quadrature coil indicated generally at 10. System 10 includes a surface coil circuit 12, indicated by a dashed enclosure, and a balun/combiner circuit 14, also indicated by a dashed enclosure. The surface coil circuit 12 includes a first coil 16 and a second coil 18. Coil 16 includes two split conductors or parallel segments 20 and 22 which are spatially displaced from each other. The inductance of each of the parallel conductor segments 20 and 22 is respectively represented by inductor 24 or 26.

Detailed Description Text (8):
Split conductors 20 and 22 are joined at node 28 to a single conductor 30 which completes the coil loop. A variable capacitor 32 is inserted in series in conductor segment 30 between nodes 34 and 36. A capacitor 38 is inserted in series between node 36 and a port 40 of the balun/computer circuit 14. Likewise, a capacitor 42 is inserted in series between node 34 and a port 44 of the balun/computer circuit 14. A fixed capacitor 46 and a variable capacitor 48 are connected between ports or nodes 40 and 44.

Detailed Description Text (9):
The inductance of coil 30 is represented at 50. Node 52 is the other terminus of split conductors 20 and 22. A capacitor 54 is connected between node 52 and a node

56. A further capacitor 58 is connected between node 52 and node 56. A variable inductor 60 is inserted in series between node 52 and a node 62. Back-to-back diodes 64 and 66 are connected in parallel between node 62 and node 56.

Detailed Description Text (10):
A fixed capacitor 68 is connected between node 56 and a node 70. In the other parallel segment or split conductor 20, a fixed capacitor 72 is connected between node 56 and a node 74. A differential capacitor indicated generally at 76 has three electrodes, two of which are connected to nodes 70 and 74, respectively, and a third of which is connected to the node 56.

Detailed Description Text (11):
Coil 18 likewise has a portion thereof split between parallel conductor segments 78 and 80 between nodes 82 and 84. The inductances of split conductors 78 and 80 are represented at 86 and 88. Fixed capacitors 90 and 92 are respectively connected in series in split conductors 78 and 80. One electrode each of capacitors 90 and 92 is connected to a node 94. A pair of fixed capacitors 96 and 98 is connected in parallel between node 84 and node 94. A pair of back-to-back diodes 100 and 102 connects node 94 to a node 104, which in turn is connected by a variable inductor 106 to the node 84.

Detailed Description Text (12):
A single conductor segment 110 is connected between junction nodes 82 and 84, and its inductance is represented at 112. A node 114 on the single conductor 110 is connected to a fixed capacitor 116, which in turn is connected to a node or port 118 of the balun/combiner circuit 14. A variable capacitor 120 is connected in series between node 114 and a node 122, both of which are on single element conductor 110. The node 122 is connected via a fixed capacitor 124 to a port or node 126 of the balun/combiner circuit 14. The balun/combiner circuit 14, the details of which are mostly unimportant here, includes a 90.degree. combiner network 128. A fixed capacitor 130 and a variable capacitor 132 span the ports 118 and 126 in parallel.

Detailed Description Text (13):
The second pair of parallel segments 78 and 80 form an overlap area with the coil 16 containing parallel segments 20 and 22. By "overlapping" or "overlap," we mean the area of projection of one coil onto the other, where both coils define areas that may or may not be substantially planar, and are positioned with respect to each other such that pan of the area of one coil has a projection onto pan of the area of the other coil. Each of the loops 16 and 18 is split into respective parallel signal path segments 20, 22 and 78, 80 located within critically overlapped area 140. Differential capacitor 76 is connected to segments 20 and 22 to differentially adjust the capacitance in series thereof, and therefore the ratio of the RF current flowing through these paths. Adjusting the relative capacitance in a differential manner allows the isolation to be adjusted without significantly changing other coil parameters such as resonant frequency or impedance matching.

Detailed Description Text (14):
A detail of the differential capacitor circuit is given in FIG. 2. In FIG. 2, the differential capacitor 76 is placed in line with parallel loop segments 20 and 22. Adjustment of the differential capacitor 76 varies the ratio of RF current flowing through parallel segments 20 and 22 of magnetic resonance coil 16.

Detailed Description Text (15):
The coils 16 and 18 (FIG. 1) are shaped in position such that the mutual inductance and therefore the majority of the coupling between the loops are minimized. This is accomplished by overlapping a critical portion of the area enclosed by the loops, such that the vector sum of all flux acting on one coil due to the other coil is near zero. However, the positioning of the loops 16 and 18 is highly critical and too sensitive to allow mass-produced devices to perform well without further individual adjustment to compensate for tolerances in physical dimensions and in values of electrical components.

Detailed Description Text (16):
To effect individual adjustment of coil system 10, an adjustment is made of differential capacitor 76 once it has been inserted into surface coil circuit 12. The results of such an individual adjustment can be measured using a network analyzer set to display the S21 (transmission) parameter. With coils 16 and 18 on a corresponding analyzer port, the coils are adjusted to minimize the value of S21.

Detailed Description Text (17):
Adjusting the capacitance in series with each of the parallel segments 20, 22 varies the ratio of the RF current flowing through each segment of the loop. This accomplishes an effect equivalent to physically moving the loops to change the amount of overlap area and shared flux. The resulting effective change in the overlapped area effectively changes the ratio of the out-of-phase flux to in-phase flux of the loops.

Detailed Description Text (18):
FIG. 3 is a schematic diagram of an alternative embodiment of the invention, further including a representative NMR quadrature coil pair 150, a pair of varactor dimes 152 and 154 coupled to coil 158, and bias circuitry for control of the electrical adjustment of the quadrature magnetic resonance surface coil pair 150. In this example, two coils 156 and 158 are shown with an overlap area 160. Each of the loops 156 and 158 is split into respective parallel signal path segments 162, 164 and 166, 168 located within a critically overlapped area 160. Varactor diodes 152 and 154 are respectively connected to segments 168 and 166 and act to adjust the capacitance thereof, and therefore the ratio of the RF current flowing through these paths.

Detailed Description Text (19):
The capacitance of varactor diodes 152 and 154 is adjusted by varying the reverse bias voltage on the varactor diodes 152 and 154. The varactor diodes 152 and 154 have a reverse orientation with respect to each other so that adjustment of the reverse bias voltage will have an opposite effect on the capacitance of each varactor diode. The reverse bias voltage is applied to varactor diodes 152 and 154 at nodes 170 and 172.

Detailed Description Text (20):
The bias voltage is supplied from DC voltage source 174 and variable resistor 176. Variable resistor 176 adjusts the capacitance of parallel signal paths 166 and 168 by varying the magnitude of reverse bias on varactor diodes 1.70 and 172. Adjustment of variable resistor 176 alters the capacitance of varactor diodes 152 and 154 in a common mode manner, changing the resonance as if a conventional variable capacitor were employed. The bias voltage is connected through RF chokes 178, 180 and 182 to prevent the loss of RF energy from the resonant circuit of the coils. DC blocking capacitors 184 and 186 prevent the DC bias voltage from creating current flow in the magnetic resonance loop 158. Fixed capacitors 188 and 190 establish a resonance condition in coil loop 156 equivalent to that in loop 158. It is clear that capacitors 188 and 190 could be replaced with a second varactor diode configuration similar to that used on loop 158.

Detailed Description Text (21):
Variable resistor 176 controls the bias voltage applied to both varactor dimes 152 and 154; it provides adjustment to the resonant frequency of the loop without changing the orthogonality adjustment established by the ratio of currents in segments 166 and 168. Variable resistor 192 is connected to the bias voltage output from variable resistor 176 and provides a secondary adjustment of the bias voltage by varying the voltage ratio between varactor dimes 152 and 154. Adjustment of variable resistor 192 varies the capacitance of varactor dimes 152 and 154 in a differential manner, allowing the isolation to be adjusted by changing the ratio of current in segments 166 and 168 without changing the resonant frequency of the coil system, as if a differential capacitor were employed.

Detailed Description Text (22):
A differential capacitor is shown generally in FIG. 2 and a functional equivalent thereof, namely a pair of varactor dimes, is shown in FIG. 4. The varactor dime pair 200, 202 of FIG. 4 is interchangeable in function with the differential capacitor circuit 76 of FIG. 2, but with the added ability to be controlled in a common mode manner also to adjust coil tuning. In FIG. 2, the differential capacitor 76 is placed in line with parallel loop segments 20 and 22. Adjustment of the differential capacitor 76 varies the ratio of RF current flowing through parallel segments 20 and 22 of magnetic resonance coil 16.

Detailed Description Text (23):
FIG. 4 shows a pair of varactor dimes 200 and 202 which are placed in line with parallel segments 204 and 206 of magnetic resonance coil 208. The capacitances of varactor dimes 200 and 202 are adjusted by varying a bias voltage 210, which is connected to varactor dimes 200 and 202 through RF dimes 212 and 214, respectively. Adjustment of the bias voltage 210 therefore adjusts the ratio of the RF current

flowing through parallel segments of magnetic resonance coil 208. Throughout the remaining FIGUREs, the varactor diode pair 200, 202 and associated bias voltage circuitry may be substituted for each differential capacitor symbol. The illustrated varactor diode pair 200, 202 could be replaced by a differential capacitor 76 of FIG. 1 in the form of a mechanical device with moving plates as actuated by an operator, a motor, or the like.

Detailed Description Text (24):
FIG. 5 shows an application of the present invention to a quad birdcage coil shown generally at 220. First and second differential capacitors 222 and 224 operate in a manner identical to that described previously; however, each differential capacitor is connected to a separate loop with respective parallel segments 226, 228 and 230, 232. Each differential capacitor may be independently tuned as previously discussed to provide the desired isolation of the coils. The differential capacitors may be formed by a pair of varactor diodes as previously described. First and second outputs are available across inductors 234 and 236, respectively.

Detailed Description Text (25):
FIG. 6 is a schematic diagram illustrating the application of the present invention to a quad multiple port birdcage coil shown generally at 238. In this example four differential capacitors 240, 242, 244, and 246 are employed to provide the desired isolation of respective coil pairs. Each of the differential capacitors is connected to respective parallel segments 248, 250; 252, 254; 256, 258; and 260, 262 to provide for the adjustment of the RF current flowing through the respective paths as previously discussed. As with the other designs it is contemplated that in one embodiment the differential capacitors be formed by pairs of varactor diodes. Output coils 264 and 266 are provided in conjunction with parallel segments 248 and 250. Output coils 268 and 270 are provided in conjunction with parallel segments 252 and 254.

Detailed Description Text (27):
FIG. 8 is a schematic diagram illustrating the present invention embodied in a multiple port planar coil shown generally at 310. The diagram illustrates a pair of differential capacitors 312 and 314 respectively located in line with separate, overlapping coils 316 and 3 18. Coil 316 is separated into parallel segments 320 and 322. Coil 318 is formed with parallel segments 324 and 326. Segments 320-326 are located within a critically overlapped area 328. The differential capacitors 312 and 314 operate in an identical manner to that previously discussed and may be formed by varactor diodes. In this embodiment differential capacitors 312 and 314 may be adjusted independently or simultaneously to provide the desired isolation of the coils.

Detailed Description Text (28):
FIG. 9 is a schematic diagram illustrating the application of the present invention to a quadrature planar multiport coil shown generally at 330. In this example two differential capacitors 332 and 334 are employed to provide the desired isolation of respective coil pairs 336 and 338. Each of the differential capacitors 332 and 334 is connected in series to respective parallel segments 340, 342 and 344, 346 to provide for the adjustment of the RF current ratio flowing through the respective paths as previously discussed. As with other designs, it is understood that more or less pairs of parallel segments in the critically overlapped areas 348, 350 can be used to increase the flexibility of orthogonality and isolation adjustment available. Also as with the other designs it is contemplated that the differential capacitors may be formed by pairs of varactor diodes.

Detailed Description Text (30):
In summary, a novel means of adjusting the orthogonality of fields generated by overlapping quadrature coils, or individual coil systems in a multicoil configuration, is disclosed. However, the above description is not intended to limit the present invention in any way, which is limited only by the scope and spirit of the following claims.

CLAIMS:

1. A magnetic resonance coil system, comprising:

a first coil;

a second coil having first and second segments configured in parallel, said second

coil disposed with respect to said first coil such that there is a physical overlap of said first and second coils; and

a differential capacitor contributing a first capacitance in series with said first conductor segment and a second capacitance in series with said second conductor segment, said differential capacitor operable to vary said first and second capacitances such that when one of said first and second capacitances is increased, the other of said first and second capacitances is reduced, thus varying the ratio of RF current in the segments first and second conductor segments.

2. The magnetic resonance coil system of claim 1, wherein said differential capacitor is remotely operable to vary said first and second capacitances.

3. The system of claim 1, wherein said differential capacitor comprises first and second varactor diodes disposed in respective ones of said first and second conductor segments, voltage circuitry coupled to apply a first voltage to said first varactor diode and a second voltage to said first and second capacitances.

5. The system of claim 3, wherein said first coil comprises first coil comprises first and second conductor segments configured in parallel, said system further comprising a second differential capacitor contributing a third capacitance in series with said first conductor segment of said first coil, said second differential capacitor contributing a fourth capacitance in series with said second conductor segment of said first coil, said second differential capacitor operable to vary said third and fourth capacitances and to thus vary the ratio of RF current in the first and second conductor segments of the second coil.

10. The system of claim 3, wherein said voltage circuitry comprises means for controlling the voltages on the varactor diodes in a differential mode to effect orthogonality adjustment and means for controlling the voltages on the varactor diodes in a common mode to effect tuning.

11. The system of claim 1, wherein said first coil has first and second conductor segments configured in parallel, all of said conductor segments of said first and second coil disposed within an overlap area.

15. A magnetic resonance imaging system, comprising:

a surface coil having first and second conductor segments connected in parallel; and

a differential capacitor having a first capacitance coupled to said first conductor segment and a second capacitance coupled to said second conductor segment, said differential capacitor being remotely controllable to differentially change said first and second capacitances such that when one of said first and second capacitances is increased, the other of said first and second capacitances is reduced.

16. A magnetic resonance imaging system, comprising:

a plurality of surface coils each having first and second conductor segments connected in parallel; and

for each of said surface coils, a differential capacitor having a first capacitance coupled to said first conductor segment of the last said surface coil and a second capacitance coupled to said second conductor segment of the last said surface coil, said differential capacitor being remotely controllable to differentially change said first and second capacitances such that when one of said first and second capacitance is increased, the other of said first and second capacitances is reduced.

17. A quadrature magnetic resonance coil system, comprising:

a first coil;

a second coil physically overlapping said first coil and having first and second conductor segments connected in parallel;

a first capacitor providing a first capacitance to said first conductor segment; and

a second capacitor providing a second capacitance to said second conductor segment, said first and second capacitors variable to adjust the current flowing in said first and second conductor segments.

18. A method for adjusting quadrature magnetic resonance surface coils, comprising the steps of:

locating first and second magnetic resonance surface coils such that said first and second coils have an area of overlap;

selectively directing the current of the first of said magnetic resonance surface coils between two parallel conductor segments of the first coil located in the area of overlap; and

varying the capacitance of said parallel conductor segments to adjust the isolation of said magnetic resonance surface coils.

19. A method for adjusting quadrature magnetic resonance surface coils, comprising the steps of:

within a critical area of overlap of first and second coils, dividing at least the first coil into first and second parallel conductor segments;

coupling a differential capacitor to the first coil such that the differential capacitor contributes a first capacitance to the first conductor segment and a second capacitance to the second conductor segment; and

varying the first and second capacitances to adjust the orthogonality of the first coil with respect to the second coil.

20. The method of claim 19, and further comprising the steps of:

forming the differential capacitor by a first varactor diode coupled to the first conductor segment and a second varactor diode coupled to the second conductor segment; and

varying voltages applied to the first and second varactor diodes to respectively adjust the first and second capacitances.

21. A quadrature magnetic resonance coil system, comprising:

a first coil;

a second coil having first and second conductor segments connected in parallel, said second coil disposed with respect to said first coil such that there is an overlap of said first and second coils; and

differential means for varying reactance contributing a first reactance to said first conductor segment and a second reactance to said second conductor segment, said differential means operable to vary said first and second reactances.

22. A magnetic resonance coil system, comprising:

a first coil;

a second coil disposed such that a principal electromagnetic field of said first coil is orthogonal to a principal electromagnetic field of said second coil, said second coil including first and second conductor segments configured in parallel; and

a differential capacitor contributing a first capacitance in series with said first conductor segment and a second capacitance in series with said second conductor segment, said differential capacitor operable to vary said first and second capacitances so as to adjust the relative orthogonality of the electromagnetic fields of said first and second coils.

23. The magnetic resonance coil system of claim 1, wherein said first and second

coils are quadrature coils.

24. The magnetic resonance coil system of claim 1, wherein said first and second coils are a portion of a phased array coil system.

L19: Entry 2 of 3 File: USPT Dec 19, 1995

DOCUMENT-IDENTIFIER: US 5477146 A
TITLE: NMR adjustable volume array

Abstract Text (1):
A NMR abdominal coil array includes four critically overlapped resonant loops fixed upon a generally "C-shaped" (curved) coil form with an anterior housing pivotally connected and supported by a posterior form. Three of the loops are fixed to an anterior housing which is curved from a generally horizontal upper end to a generally vertical lower end, and pivotally connected to a generally vertically oriented flange forming the upper end of the posterior housing. This upper end of the posterior housing contains the remaining resonant loop. The anterior housing is pivotally connected to the posterior housing about a single horizontal axis, permitting both slight adjustment for patient size, and substantial movement for entry and exit of a patient.

Brief Summary Text (2):
The present invention relates generally to magnetic resonance imaging (MRI) and more particularly to local coils for use in receiving MRI signals.

Brief Summary Text (4):
A. Magnetic Resonance Imaging

Brief Summary Text (5):
Magnetic resonance imaging (MRI) refers generally to a form of clinical imaging based upon the principles of nuclear magnetic resonance (NMR). Any nucleus which possesses a magnetic moment will attempt to align itself with the direction of a magnetic field, the quantum alignment being dependent, among other things, upon the strength of the magnetic field and the magnetic moment. In MRI, a uniform magnetic field B.sub.0 is applied to an object to be imaged; hence creating a net alignment of the object's nuclei possessing magnetic moments. If the static field B.sub.0 is designated as aligned with the z axis of a Cartesian coordinate system, the origin of which is approximately centered within the imaged object, the nuclei which posses magnetic moments precess about the z-axis at their Larmor frequencies according to their gyromagnetic ratio and the strength of the magnetic field.

Brief Summary Text (6):
Water, because of its relative abundance in biological tissues and its relatively strong net magnetic moment M.sub.z created when placed within a strong magnetic field, is of principle concern in MR imaging. Subjecting human tissues to a uniform magnetic field will create such a net magnetic moment from the typically random order of nuclear precession about the z-axis. In a MR imaging sequence, a radio frequency (RF) excitation signal, centered at the Larmor frequency, irradiates the tissue with a vector polarization which is orthogonal to the polarization of B.sub.0. Continuing our Cartesian coordinate example, the static field is labeled B.sub.z while the perpendicular excitation field B.sub.1 is labeled B.sub.xy. B.sub.xy is of sufficient amplitude and duration in time, or of sufficient power to nutate (or tip) the net magnetic moment into the transverse (x-y) plane giving rise to M.sub.xy. This transverse magnetic moment begins to collapse and re-align with the static magnetic field immediately after termination of the excitation field B.sub.1. Energy gained during the excitation cycle is lost by the nuclei as they re-align themselves with B.sub.0 during the collapse of the rotating transverse magnetic moment M.sub.xy.

Brief Summary Text (7):
The energy is propagated as an electromagnetic wave which induces a sinusoidal

signal voltage across discontinuities in closed-loop receiving coils. This represents the NMR signal which is sensed by the RF coil and recorded by the MRI system. A slice image is derived from the reconstruction of these spatially-encoded signals using well known digital image processing techniques.

Brief Summary Text (8):
B. Local Coils and Arrays

Brief Summary Text (9):
The diagnostic quality or resolution of the image is dependent, in part, upon the sensitivity and homogeneity of the receiving coil to the weak NMR signal. RF coils, described as "local coils" may be described as resonant antennas, in part, because of their property of signal sensitivity being inversely related to the distance from the source. For this reason, it is important to place the coils as close to the anatomical region-of-interest (ROI) as possible.

Brief Summary Text (10):
Whereas "whole body" MRI scanners are sufficiently large to receive and image any portion of the entire human body, local coils am smaller and therefore electromagnetically couple to less tissue. Coupling to less tissue gives rise to coupling to less "noise" or unwanted biologically or thermally generated random signals which superimpose upon the desired MR signal. The local coils may be of higher quality factor (Q) than the body coils due to their smaller size. For all of these reasons, local coils typically yield a higher signal-to-noise (S/N) ratio than that obtainable using the larger whole body antenna. The larger antenna is commonly used to produce the highly homogenous or uniform excitation field throughout the ROI, whereas the local coil is placed near the immediate area of interest to receive the NMR signal. The importance of accurate positioning leads to the development of local coils which conform to the anatomy of interest, yet function to permit ease of use.

Brief Summary Text (11):
While the smaller local coil's size works to an advantage in obtaining a higher S/N ratio, this reduced size also presents a disadvantage for imaging deep-seated tissues. Typically, the single- conductor coil diameter which yields the optimal S/N ratio at a depth `d` is a coil of diameter `d`; hence, larger diameter single-conductor coils are required to image regions in the abdomen and chest of human patients. This increased coil size results in less than desirable performance, both in terms of S/N ratio and homogeneity of the sensitivity profile (which effects the uniform brightness of the image), and offers little advantage over the body coil of the system.

Brief Summary Text (12):
The S/N ratio of the NMR signal may be further increased within a region by digitally adding the post processed signals derived from two or more coils; each sensitive to the precessing nuclei within a common volume. If two coils' signals are processed and converted into image data separately and then added digitally, one can obtain an increase in S/N ratio within the common volume without the use of a signal combiner. Separate amplifiers, analog-to-digital converters, and image processor channels represent an alternative configuration for processing two signals in lieu of a single combiner circuit and processing channel. Such a system of four channels whose signals are derived from an array of four coils is described in U.S. Pat. No. 4,825,162. In the '162 patent, an array of coils is described wherein the adjacent coils overlap to prevent nearest neighbor interaction (inductive coupling). The interaction between the next nearest neighbor is supposedly reduced by connection of each coil of the array to low input impedance preamplifiers.

Brief Summary Text (13):
The problem with this solution is, among other things, the use of preamplifiers with low input impedance. This additional circuitry is costly and adds another set of possible failure modes into the system. This preamplifier circuitry is sensitive to coil impedance changes resulting from patient loading variations as well as to noise spikes or power surges within the receiver chain.

Brief Summary Text (14):
One can minimize the effects of next-nearest-neighbor interaction if one properly utilizes the formulation, in the following arguments, to minimize inductive coupling between all resonant structures. In this case, the additional preamplifier circuitry is no longer required. First, nearsest--neighbor or adjacent coil interaction is a

much more dominant coupling than the next-nearest-neighbor coupling--usually one or two orders of magnitude larger depending upon coil size and spacing. Second, if near-neighbor coupling has not been sufficiently minimized, then next-nearest neighbor coupling will occur via neighbor-to-neighbor interaction as strongly as, or stronger than inductive coupling between next-nearest-neighbors. Third, next-nearest neighbor interaction (inductive coupling) is further reduced towards zero when the next-nearest-neighbor coils are dominantly loaded by coupling to patient tissues. Such is the case in mid to high field scanners operating above 20 MHz. The coil's impedance is also dominated resulting from coupling to eddy current loops generated within the patient tissues. This is predicted from the mutual impedance formulation ##EQU1## where Z.sub.1d is the driving or output impedance of coil 1, Z.sub.11 is the self-impedance of coil 1, (I.sub.2 /I.sub.1) is the ratio of induced eddy currents (loop 2) to the current in coil 1, and Z.sub.12 is the mutual impedance between the loops which is equal to the radian frequency times the mutual inductance between said loops.

Brief Summary Text (15):
The implication from the above three facts is as follows. If one ensures consistent and dominant loading of the coil elements and if one ensures that near-neighbor coupling has been minimized (that inductive isolation has been achieved) and if the antenna element size, geometrical orientation, and spacing is designed so as to minimize next-nearest neighbor coupling, then the array will work properaly with little degrading interaction amongst the elements.

Brief Summary Text (16):
Inductive isolation is achieved by geometrically orienting two coil conductors such that their mutual inductance is minimized according to the following: ##EQU2## where M represents the mutual inductance between coils 1 and 2 and the vector components dl.sub.1 and dl.sub.2 represent segments of coils 1 and 2 with current amplitudes I.sub.1 and I.sub.2. The denominator represents the magnitude difference of the position vectors of each dl segment. The condition wherein M is approximately zero with respect to the individual self inductances of coils 1 and 2, is known as geometric isolation between the coils. This is a special case of inductive isolation but is restrictive in application, as discussed below.

Brief Summary Text (17):
As the coil geometries are sufficiently large or close to the surrounding system conductors (antenna, faraday screen, cryostat tubing, etc.) in addition to the biological conducting medium, this coupling formula must be extended to include M=M.sub.12 +M.sub.13 +M.sub.23 ; where the first term is as described above, and the latter two terms define the coupling resulting from each coil's coupling to eddy current loops (loop 3) generated on or within the surrounding conductors (system or biological). These additional coupling terms must be accounted for in the adjustment of conductor geometries with respect to each other spatially. With these terms taken into account, the proper critical spacing may be found between coil loops 1 and 2.

Brief Summary Text (18):
An additional problem is not fully addressed by the prior art: this is the challenge to optimize image quality from deep seated tissues which cannot practically be surrounded by a coil conductor geometry of any appreciably reduced size in comparison to the existing body coil. This tissue geometry therefore warrants some type of surface coil configuration. An example of this would be the region of the chest or abdomen. Due to physical size limitations of the magnetic resonance imaging system bore, there is no room available to place a typical patient into an inner cylindrical volume coil such as the birdcage design of U.S. Pat. Nos. 4,680,548 and 4,692,705.

Brief Summary Text (19):
The early quadrature coil patent (U.S. Pat. No. 4,866,387) addressed the geometric isolation problem between two resonant structures but did not address the practical issues of implementation: signal combination without destruction of individual coil tuning; optimization of coil sensitivity to deep-seated tissues; and maximization of coil efficiency through minimization of coil effective volume. Quadrature signal combination cannot be accomplished without either suppressing shunt RF currents which exist through paths through the combiner circuitry, or re-tuning each coil element to compensate for the shunt current's detuning effects. Often the latter is not possible, and when it is, any frequency shift due to loading variations from patient to patient render this solution useless and quadrature performance is not achieved.

Brief Summary Text (20):
Coil efficiency is the magnetic field per unit current (upon the conductor) and provides an indication as to the ability to resolve small signals from a noisy environment. The effective volume of the coil is a calculated ratio of the coil's total region-of-sensitivity (ROS) (volume in which the coil can distinguish signal from noise) divided by its largest efficiency value within the center of the coil (not directly adjacent to any conductor). The resulting numeric value is useful for optimizing the coil geometry with respect to sensitivity from the region of interest. Sensitivity is improved with coupling to fewer noise sources, and the body represents a large volume of biologically created noise sources; therefore, if the coil is designed to couple to less tissue where signal is not desired, and its magnetic field profile is focused on tissues where signal is desired (optimal effective volume), then a sensitivity and efficiency performance will be realized.

Brief Summary Text (21):
Phased array coils such as described in U.S. Pat. Nos. 4,825,162 and 5,198,768 also do not address the problem of obtaining the optimal image quality from deep-seated tissues as discussed above. Both patents focus on obtaining a larger ROS using a bank of coils whose signals input to separate preamplifiers and digital reconstruction ports on a computer. Neither patent addresses the need for designing each coil configuration for the purpose of overlapping each coil's ROS within the desired deep seated volume and optimizing each with respect to performance within that region as discussed above. The aforementioned prior art teaches linear coil array technology--overlapping coils which extend the ROS to a larger region with the performance of a smaller coil within the entire region.

Brief Summary Text (22):
Finally, the prior art is reliant upon geometric isolation and/or low impedance preamps with no compensation for eddy current-induced coupling, and restricts that geometry to a planar surface only. The prior art also does not address imaging within a common volume by utilizing an array of coils connected to a set of separate receiver channels. This restriction is due to the fact that the prior art is dependent upon geometric isolation only, and this alone is inadequate to ensure sufficient isolation between non-adjacent pairs of non-planar conductors.

Brief Summary Text (23):
The prior art also does not address the design complexities of imaging within a common volume which varies in size or girth using coil array technology.

Brief Summary Text (25):
It is an object of the present invention to provide an improved NMR local coil designed to conform to a patient's torso within the geometric restriction of the surrounding system bore, and place multiple antenna conductors within close proximity to the entirety of said anatomy.

Brief Summary Text (27):
Still another object is to provide a set of non-planar coil conductors which closely couple their regions of sensitivity to a patient's torso without detuning each individual resonator due to coupling between said resonators.

Brief Summary Text (28):
Another object of the present invention is to provide improved electronic configurations of coil conductors which yield a higher signal-to-noise ratio and improved homogeneity (of sensitivity profiles) within a deep-seated volume of interest.

Brief Summary Text (29):
A further object is to provide a co-located coil set which will interface with either a multiple-channel receiver system or a single-channel receiver system and yield an improved S/N ratio within a common volume of interest which can vary in size or girth.

Brief Summary Text (31):
The NMR abdominal coil array of the present invention includes four critically overlapped resonant loops fixed upon a generally "C-shaped" (curved) coil form with an anterior housing pivotally connected and supported by a posterior form. Three of the loops are fixed to an anterior housing which is curved from a generally horizontal upper end to a generally vertical lower end, and pivotally connected to a

generally vertically oriented flange forming the upper end of the posterior housing. This upper end of the posterior housing contains the remaining resonant loop. The anterior housing is pivotally connected to the posterior housing about a single horizontal axis, permitting both slight adjustment for patient size, and substantial movement for entry and exit of a patient.

Brief Summary Text (32):
The three loops comprising the anterior conductor set provide imaging coverage laterally across the torso of the patient as well as along one side. The fourth loop is positioned such that inductive isolation is effectively maintained throughout the adjustment range of the pivotal anterior housing and coils.

Brief Summary Text (33):
An alternate configuration of electronically operating the same four loops is to connect alternating or non-adjacent pairs in Helmholtz-style configurations. This configuration operates whereby the two capacitively connected loops resonate at a frequency where strong magnetic flux coupling occurs between the pair of coils. This configuration is known amongst those skilled in the art as being useful for imaging more deep-seated tissues. Due to the geometric placement of the four loops, configuring the two Helmholtz-style pairs creates two vector magnetic field sensitivities orthogonal to each other within the common volume; hence creating a quadrature antenna set.

Brief Summary Text (34):
The coil housing facilitates close coupling of all four resonators to the desired anatomical location whereby all resonators contribute to the total signal collection from deep-seated tissues. The hinged opening allows adjustment to ensure nearly constant spacing between the coil conductors and the patient; hence constant coil loading, impedance matching, and therefore performance. This criteria is typically easy to achieve with linear arrays and single coil elements.

Drawing Description Text (2):
FIG. 1 is an exploded perspective view of the NMR adjustable volume array of the present invention, with conductor profiles superimposed thereon;

Drawing Description Text (3):
FIG. 2 is a side elevational schematic view taken from the right side of FIG. 1 with the resulting magnetic flux isocontours of each coil element (shown in the fully closed operational position) superimposed thereon;

Drawing Description Text (7):
FIG. 6 is a perspective view of the conductors and electrical connections of a second embodiment of the NMR adjustable volume array.

Detailed Description Text (2):
Referring now to the drawings, in which similar or corresponding parts are identified with the same reference numeral, and more particularly to FIG. 1, the NMR adjustable volume array of the present invention is designated generally at 10 and includes an upper anterior segment 12 operably mechanically connected to an opposed lower posterior segment 14.

Detailed Description Text (3):
Anterior segment 12 includes a plurality of electronic NMR coils 16, 18, and 20 enclosed within a durable plastic housing 22 (shown cut-away on the top side). Housing 22 extends from a generally horizontal position at upper end 24 to a generally vertical position at lower end 26, when anterior segment 12 is in the operable position (A) shown in FIG. 2.

Detailed Description Text (4):
Lower end 26 of housing 22 projects downwardly generally parallel to the generally upwardly extended flange 28 of the posterior segment 14. Flange 28 includes an electronic NMR coil conductor 30 which serves as the fourth of the array of four conductors. Conductor 30 is enclosed within a durable plastic housing 29 (shown cut-away on the bottom side). Lower end 26 of housing 22 is formed to cream an overlap between conductors 16 and 30, and serve as part of a hinge assembly 32, 34, 36, and 38 between the anterior and posterior segments, 12 and 14 respectively.

Detailed Description Text (6):
An electrical cable assembly 44 extends out from housing 29 from the flange 28, and

enters housing 22 in the lower end 26 through a hole 45. The NMR signals from coils 16, 18, and 20 are transmitted via miniature coaxial conductors 46, 48, and 50 (respectively) which are bundled together (but electrically insulated from one another) within cable assembly 44. These cables are then bundled with coil 30's transmission line, cable 52, within housing 29, to form a larger cable bundle assembly 54 which passes out of coil housing 29 via port hole 56. Cable assembly 54 serves to carry the NMR signals from all coil elements to their respective preamplifier circuitry (within the NMR system), and also carry the direct current source signals from the NMR system to the respective coil elements to activate their recoupling circuitry located on electronic circuit boards 58, 60, 62, and 64.

Detailed Description Text (7):
Referring now to FIG. 3, the electrical schematic of circuit board 58 is shown. Since the component scheme is identical to that of boards 60, 62, and 64, only circuit board 58 will be discribed in detail. Series capacitors 66 and 68 are sized to appropriately resonate the coil loop 30. This series configuration reduces the total impedance across the capacitor 68 which must be impedance matched to the 50 ohm transmission line 52. Inductors 70,72,74, and 76 and capacitor 78 form a "modified Tee" impedance matching network which matches the complex impedance developed across capacitor 68 to the 50 ohm transmission line 52. Inductors 70 and 74 are series connected between capacitors 66 and 68 and one conductor 80 of transmission line 52. Inductors 72 and 76 are series connected between the opposite side of capacitor 68 and a second conductor 82 of transmission line 52. One terminal of capacitor 78 is connected between inductors 70 and 74, while the other terminal of capacitor 78 is connected between inductors 72 and 76, to form the "modified Tee". Whereas inductors 70 and 74 in conjunction with capacitor 78 would be the "standard Tee" configuration, the additional inductors 72 and 76 modify the standard Tee and serve two purposes. First, they are designed to be approximately equal in inductance to inductors 70 and 74; hence creating a balanced-to-unbalanced impedance transformer ("balun" to those versed in radio frequency (RF) electronics). Second, dividing the total circuit inductance evenly onto both conductors 80 and 82 of the transmission line 52 also keeps both nodes 84 and 86 above earth ground so as to prevent establishment of an RF ground within the sensitive receiver system. Such an RF ground can produce undesirable effects upon the magnetic field homogeneity within the NMR system.

Detailed Description Text (8):
Diode 88 is connected at one end between inductors 70 and 74, and at the other end to node 86, and serves as a decoupling diode which is activated by system-provided direct current (DC)--voltage pulses. During the transmit mode of the MRI data acquisition cycle, a DC voltage forward biases diode 88 into a conduction state; hence, effectively placing inductor 70 in parallel with capacitor 68. Together these components create a high impedance circuit to the RF currents induced upon the resonant coil structure, loop 30, thereby decoupling the coil loop from the transmit antenna power.

Detailed Description Text (9):
Referring now to FIG. 4, the simplified conductor diagram of the coil elements 16, 18, 20 and 30, demonstrates that a critical overlap exists between the conductors such that they operate in conjunction with one another in the manner described below.

Detailed Description Text (10):
Each coil is inductively isolated from the adjacent coil via minimization of total mutual inductance M which includes the inductance between two adjacent coils M.sub.12 and the inductance between each of these coils and a third eddy current loop (loop 3--terms M.sub.13 and M.sub.23) which is created by the induction of currents upon adjacent conducting materials such as; the RF screen (not shown)--surrounding the entire assembly 10, or biological medium 79.

Detailed Description Text (11):
Referring once again to FIG. 2, the coils 16,18,20, and 30 (whose conductors are shown in position on the axial slice of the coil geometry) have been sized such that their region(s)-of-sensitivity (shown in dashed lines) penetrate well into the patient 11 but not through the patient where a significant mutual coupling would exist between non-adjacent pairs of resonant loops. Note that the iso-flux (magnetic field strength per unit current (H/I)) contours 21a, 19a, 17a, and 31a, associated with coil elements 20, 18, 16, and 30 respectively, do overlap within the central region of the patient 11, but with a H/I value (0.5) significantly reduced from the

H/I values (4.5) 21b, 19b,17b, and 31b obtainable proximal to the coil conductors. Optimization of coil elemental size has been accomplished using the above-mentioned computational algorithms so that all coil elements receive MRI signal from deep-seated tissues and yet couple minimally to one another. Signal from all coil elements are digitally sampled and added in the MRI system signal processor. Also, the distance separating the non-adjacent coil loops as well as the direction of induced currents serve to minimize coupling between the four loops in this invention. This is important for achieving the maximum performance from each coil element in terms of that element remaining impedance matched at, and tuned to, the proper frequency as discussed in detail in conjunction with FIG. 5, herein below.

Detailed Description Text (12):
Continuing with FIG. 2, the patient load (resistance coupled into the coil circuit via mutual impedance between the coil and current loops within the patient) remains relatively constant in this invention due to the hinged anterior segment 12. Anterior segment 12 houses coil elements 16, 18, and 20 and adjusts to maintain a relatively fixed distance from the patient 11 over a broad range of patient sizes. As the patient load is the dominant source of coil loading at the coil frequencies of which this coil set is operating, maintaining the fixed patient load permits limited adjustment of the coil positions relative to one another without destroying the inductive isolation which was sought in the original design concept.

Detailed Description Text (13):
Motion of coil 16 with respect to coil 30 is restricted such that mutual inductance between coils 16 and 30 remains close to zero. The small variation of mutual inductance from null is so slight that the resulting frequency shift of coils 16 and 30 also remains small. In the instance where the mutual inductance M does not equal zero, the resulting frequencies are ##EQU3## where L and C represent the total coil inductance and capacitance of each loop.

Detailed Description Text (14):
Referring now to FIG. 5, the frequency response of a coil is represented by the coil sensitivity (H/I) plotted with respect to frequency. Frequency response curve A represents a single coil loop response with peak sensitivity at the initial resonance f.sub.o.

Detailed Description Text (15):
An acceptable frequency shift due to non-zero mutual inductance is represented by trace B. At this shift the H/I is only decreased by 1 dB from maximum. This translates to a visually imperceptible change in image brightness in a processed MR image. Comparatively, a larger frequency shift such as represented by curve C results in a signal degradation of 6 dB which translates to one quarter of the signal intensity of the originally tuned coil.

Detailed Description Text (16):
Optimizing the coil sizes with respect to desired magnetic field sensitivity penetration, minimizing inductive coupling by accounting for eddy current loops as well as adjacent coil loops, and maintaining the dominant coil loading during exams of a variety of patient sizes are the procedural essence of this invention.

Detailed Description Text (17):
Referring now to FIG. 6, a second embodiment of the NMR adjustable volume array relies upon the electrical connection of coils 16, 18, 20, and 30. Non-adjacent coils 16, 20 and 18, 30 are paired such that a common magnetic flux linkage is created between the two coils of each pair. Junction capacitors 85, 87, 89, and 90 of the four coils are part of the total distributed capacitance of each coil loop 20, 16, 18, and 30 respectively.

Detailed Description Text (18):
Coil 20 is capacitively connected to non-adjacent coil 16 via a conductor pair 92 which connects one terminal of capacitor 85 to the respective terminal of capacitor 87. Opposite terminals of capacitors 85 and 87 are also connected via the remaining conductor of the conductor pair 92. This connection ensures a commonly directed current flow in each of the proximal conducting rods 96 and 98 of coils 20 and 16 respectively hence ensuring a common magnetic flux linkage between coils 20 and 16. This is similar to that which exists in a Helmholtz configuration of two coil loops such as familiar to those learned in the art.

Detailed Description Text (19):

Similarly, coils 18 and 30 are connected across capacitors 89 and 90 with conductor pair 94 creating a common magnetic flux linkage.

Detailed Description Text (20):
Referring again to FIG. 4, the common magnetic flux vector direction created with the patient imaging volume by coil pair 16 and 20 is depicted by vector 17 while vector 19 represents that flux linkage between coils 18 and 30. These vectors are generally perpendicular to each other thereby ensuring magnetic isolation between the two pairs of coils. The contour 81 indicates the region of optimal quadrature gain resulting from orthogonality and amplitude equality of vectors 17 and 19. Quadrature gain within this contral region contributes to the larger region of uniformity 83. Note that is is the combination of signal strength, which decays as the inverse of the distance squared from the coil conductors, and quadrature gain which yields this uniform region of sensitivity 83.

Detailed Description Text (21):
Referring again to FIG. 6, NMR signals generated across the capacitor pairs 85-87 and 89-90 are then either combined via a quadrature combiner circuit 100 with the resulting signal transmitted via a single transmission line 102, or the two signals may feed into separate receiver channel preamplifier circuits as discussed in the first embodiment. Each signal generated across the above mentioned capacitor pairs is impedance matched using lumped element matching circuit boards 95 and 97 and transmitted to the combiner 100 (as shown) via short lengths of coaxial cable 99 and 101, or after impedance matching the cables 99 and 101 are extended to the system preamplifier connector.

Detailed Description Text (22):
Whereas the invention has been shown and described in connection with the preferred embodiments thereof, it will be understood that many modifications, substitutions and additions may be made which are within the intended broad scope of the appended claims. There has therefore been shown and described an improved NMR local coil which accomplishes at least all of the above stated objects.

CLAIMS:

1. A NMR adjustable volume array, comprising:

an anterior housing supporting an anterior coil set, having opposing forward and rearward ends, an upper end and a lower end;

said anterior housing curved from a generally vertically oriented lower end to a generally horizontally oriented upper end when the anterior housing is in a working position;

said anterior coil set including a plurality of NMR coils maintained in fixed positions relative to one another within said housing, and following the curvature of said anterior housing;

a posterior housing supporting a posterior NMR coil, having opposing forward and rearward edges, an upper end and a lower end, operably and pivotally connected along its upper end to said anterior housing lower end such that said anterior housing is pivotable within a predetermined pivot range;

said posterior housing curved upwardly from a generally horizontally oriented lower end to a generally vertically oriented upper end;

said anterior and posterior housings forming a generally C-shaped housing assembly; a posterior NMR coil within said posterior housing and following the curvature of said housing from the upper end towards the lower end; and

said anterior coil set and posterior coil electrically connected to NMR imaging apparatus and operable throughout the pivot range of said anterior housing.

2. The array of claim 1, wherein said anterior NMR coils includes an upper coil extending from the anterior housing upper end towards the anterior housing lower end, and a lower coil extending from the anterior housing lower end towards the anterior housing upper end, and wherein said posterior and anterior coils each have a region of sensitivity extending generally radially from each said coil inwardly towards the general center of said C-shaped housing assembly.

3. The array of claim 2, further comprising means for pivotally connecting said anterior and posterior housings so as to maintain relative inductive isolation between the anterior lower coil and the posterior coil.

4. The array of claim 3, wherein each said coil is a loop antenna having an upper end and a lower end, and wherein relative inductive isolation between the anterior lower coil and the posterior coil is maintained by pivotally connecting the anterior and posterior housing with the lower end of the anterior lower coil overlapping the upper end of the posterior coil such that the posterior coil upper end projects upwardly beyond the lower end of the anterior lower coil a predetermined distance and such that the anterior housing is pivoted generally about the lower end of the anterior lower coil.

5. The array of claim 4, wherein said anterior coils include an intermediate NMR coil located substantially between said upper and lower anterior coils, and wherein said anterior coils are located so as to maintain relative inductive isolation.

6. The coil array of claim 1, wherein said anterior coil set includes an upper, an intermediate and a lower anterior coil, and further comprising:

means for electrically connecting said upper and lower anterior coils to operate as a first coil pair; and

means for electrically connecting said intermediate anterior coil and said posterior coil to operate as a second coil pair.

7. The coil array of claim 6, wherein said coils are sized and located relative to one another so as to minimize inductive coupling between said first and second coil pairs.

8. The coil array of claim 7, wherein said coils of said first coil pair are located relative to one another, and said coils of said second coil pair are located relative to one another, to minimize inductive coupling between said first and second coil pairs.

9. The coil array of claim 8, wherein said first coil pair is located relative to the second coil pair to minimize inductive coupling therebetween.

10. The coil array of claim 9, wherein each said first and second coil pairs have a magnetic field vector, and wherein said first and second coil pairs are located to orient said vectors orthogonally.

11. The coil array of claim 6, further comprising means for electrically connecting the coils of said first coil pair to create a Helmholtz pair, and means for electrically connecting the coils of said second coil pair to create a second Helmholtz pair.

12. The coil array of claim 6, wherein each said coil pair detects precessing magnetic moment in tissue produced by NMR imaging apparatus and transmits a signal in response thereto, and wherein said electrical connection to the NMR imaging apparatus further comprises independent transmission lines connected between each coil pair and independent receiver preamplifiers in said NMR imaging apparatus.

13. The coil array of claim 6, wherein each said coil pair detects precessing magnetic moment in tissue produced by NMR imaging apparatus and transmits a signal in response thereto, wherein said electrical connection to the NMR imaging apparatus further comprises:

independent transmission lines connected between each coil pair and a combiner means for combining said signals in quadrature; and

a single transmission line connected between said combiner means and a single receiver preamplifier in the NMR imaging apparatus.

End of Result Set

Generate Collection | Print

L19: Entry 3 of 3 File: USPT Sep 15, 1987

DOCUMENT-IDENTIFIER: US 4694255 A
TITLE: Radio frequency field coil for NMR

Abstract Text (1):
An NMR radio-frequency coil is made up of a plurality of conductive segments evenly spaced about the peripheries and interconnecting a pair of conductive loop elements. Each conductive segment includes at least one reactive element which may include a variable capacitive or inductive element.

Brief Summary Text (2):
This invention relates to nuclear magnetic resonance (NMR) apparatus. More specifically, this invention relates to radio frequency (RF) coils useful with such apparatus for transmitting and/or receiving RF signals.

Brief Summary Text (3):
In the past, the NMR phenomenon has been utilized by structural chemists to study, in vitro, the molecular structure of organic molecules. Typically, NMR spectrometers utilized for this purpose were designed to accommodate relatively small samples of the substance to be studied. More recently, however, NMR has been developed into an imaging modality utilized to obtain images of anatomical features of live human subjects, for example. Such images depicting parameters associated with nuclear spins (typically hydrogen protons associated with water in tissue) may be of medical diagnostic value in determining the state of health of tissue in the region examined. NMR techniques have also been extended to in vivo spectroscopy of such elements as phosphorus and carbon, for example, providing researchers with the tools, for the first time, to study chemical processes in a living organism. The use of NMR to produce images and spectroscopic studies of the human body has necessitated the use of specifically designed system components, such as the magnet, gradient and RF coils.

Brief Summary Text (4):
By way of background, the nuclear magnetic resonance phenomenon occurs in atomic nuclei having an odd number of protons and/or neutrons. Due to the spin of the protons and neutrons, each such nucleus exhibits a magnetic moment, such that, when a sample composed of such nuclei is placed in a static, homogeneous magnetic field, B.sub.o, a greater number of nuclear-magnetic moments align with the field to produce a net macroscopic magnetization M in the direction of the field. Under the influence of the magnetic field B.sub.o, the magnetic moments precess about the axis of the field at a frequency which is dependent on the strength of the applied magnetic field and on the characteristics of the nuclei. The angular precession frequency, .omega., also referred to as the Larmor Frequency, is given by the equation .omega.=.gamma.B, in which .gamma. is the gyromagnetic ratio (which is constant for each NMR isotope) and wherein B is the magnetic field (B.sub.o plus other fields) acting upon the nuclear spins. It will be thus apparent that the resonant frequency is dependent on the strength of the magnetic field in which the sample is positioned.

Brief Summary Text (5):
The orientation of magnetization M, normally directed along the magnetic field B.sub.o, may be perturbed by the application of magnetic fields oscillating at or near the Larmor frequency. Typically, such magnetic fields designated B.sub.1 are applied orthogonal to the direction of magnetization M by means of radio-frequency pulses through a coil connected to radio-frequency-transmitting apparatus. Magnetization M rotates about the direction of the B.sub.1 field. In NMR, it is

typically desired to apply RF pulses of sufficient magnitude and duration to rotate magnetization M into a plane perpendicular to the direction of the B.sub.o field. This plane is commonly referred to as the transverse plane. Upon cessation of the RF excitation, the nuclear moments rotated into the transverse plane begin to realign with the B.sub.o field by a variety of physical processes. During this realignment process, the nuclear moments emit radio-frequency signals, termed the NMR signals, which are characteristic of the magnetic field and of the particular chemical environment in which the nuclei are situated. The same or a second RF coil may be used to receive the signals emitted from the nuclei. In NMR imaging applications, the NMR signals are observed in the presence of magnetic-field gradients which are utilized to encode spatial information into the NMR signal. This information is later used to reconstruct images of the object studied in a manner well known to those skilled in the art.

Brief Summary Text (6):
In performing whole-body NMR studies, it has been found advantageous to increase the strength of the homogeneous magnetic field B.sub.o. This is desirable in the case of proton imaging to improve the signal-to-noise ratio of the NMR signals. In the case of spectroscopy, however, this is a necessity, since some of the chemical species studied (e.g., phosphorus and carbon) are relatively scarce in the body, so that a high magnetic field is necessary in order to detect usable signals. As is evident from the Larmor equation, the increase in magnetic field B is accompanied by a corresponding increase in .omega. and, hence, in the resonant frequency of the transmitter and receiver coils. This complicates the design of RF coils which are large enough to accommodate the human body. One source of difficulty is that the RF field generated by the coil must be homogeneous over the body region to be studied. Another complication arises from the intrinsic distributed inductance and capacitance in such large coils which limit the highest frequency at which the coil can be made to resonate.

Brief Summary Text (7):
Presently used coils employ one turn or two turns in parallel to minimize the inductance and increase the resonant frequency. The concentration of the resonant current in so few turns reduces the homogeneity of the B.sub.1 field and homogeneity of the sensitivity to signals produced in different parts of the sample region. Moreover, the lack of symmetry between the position of the tuning capacitor and the stray capacitance of the single-turn coil lead to a non-uniform current distribution in the coil and a corresponding reduction in the uniformity of the B.sub.1 field and signal sensitivity.

Brief Summary Text (9):
It is another object of the invention to provide an NMR RF coil which is operable at lower RF power and which exhibits an improved signal-to-noise ratio.

Brief Summary Text (10):
It is still another object of the invention to provide an NMR RF coil having current and tuning capacitance distributed in many turns but which has an effective inductance of a single turn.

Brief Summary Text (12):
In accordance with the invention, an NMR radio-frequency-field coil includes a pair of conductive elements disposed in a spaced-apart relation along a common longitudinal axis. The loop elements are electrically interconnected by a plurality of conductive segments each having at least one reactive element in series therewith. The segments are disposed substantially parallel to the common longitudinal axis. In one embodiment, the segments are spaced along the loop peripheries such that the resulting configuration has four-fold symmetry. In another embodiment the segments are spaced such that the resulting geometry does not have four-fold symmetry.

Drawing Description Text (3):
FIG. 1A illustrates in schematic form a conventional, parallel-connected, two-turn NMR RF coil used for whole-body studies;

Drawing Description Text (5):
FIG. 1C depicts in schematic form another conventional two-turn, series-connected NMR RF coil used in NMR studies of the head, for example;

Drawing Description Text (6):

FIG. 1D depicts as yet another conventional NMR RF coil;

Drawing Description Text (7):
FIG. 2A depicts a single-turn saddle coil which forms the basic element of the coil constructed in accordance with the invention;

Drawing Description Text (8):
FIG. 2B depicts schematically the inventive NMR RF coil;

Drawing Description Text (9):
FIG. 3A is a lumped-element equivalent circuit of the inventive NMR RF coil;

Drawing Description Text (11):
FIG. 4 depicts an embodiment of the inventive NMR RF coil capable of being driven at two points;

Drawing Description Text (12):
FIG. 5A depicts a top view of an inventive NMR RF coil illustrating current direction in the conductive loop element;

Drawing Description Text (13):
FIG. 5B depicts the sinusoidal current distribution in the vertical segments of the coil depicted in FIG. 5A;

Drawing Description Text (14):
FIG. 5C depicts the current distribution in the conductive loop element of the coil depicted in FIG. 5A;

Drawing Description Text (15):
FIG. 5D illustrates an inventive NMR RF coil having a window formed therein and including a deformed conductive loop lement;

Drawing Description Text (16):
FIG. 5E depicts the current distribution in the vertical segments for the inventive coil depicted in FIG. 5D having some segments removed;

Drawing Description Text (17):
FIGS. 6A-6C are similar to FIGS. 5A, 5B, and 5C, respectively, and depict an inventive NMR RF coil, and currents associated therewith, having open circuits formed in the loop elements thereof;

Drawing Description Text (18):
FIGS. 7A and 7B depict conductive patterns utilized in fabricating the preferred embodiment of the inventive NMR RF coil;

Drawing Description Text (19):
FIG. 8A depicts one embodiment of the inventive NMR RF coil made up of a plurality of vertical rods with a variable capacitor built into each rod;

Drawing Description Text (21):
FIG. 9A shows another embodiment of the inventive NMR RF coil which is made up of a plurality of vertical rods with fixed capacitors at each end;

Drawing Description Text (23):
FIG. 10 depicts as yet another embodiment of the inventive NMR RF coil comprised of a plurality of parallel wires shorted together at one end and plated onto the outside surfaces of a telescoping form comprised of dielectric material;

Drawing Description Text (24):
FIG. 11 depicts still another embodiment of the inventive NMR RF coil made up of a plurality of parallel isolated wires which are plated onto the outside surface of a cylinder formed of dielectric material; and

Drawing Description Text (25):
FIG. 12 is a partial schematic illustration of the inventive NMR RF coil in which selected ones of the vertical wires are provided with a plurality of fixed capacitors which may be utilized for impedance matching.

Detailed Description Text (2):

A solenoidal geometry is frequently utilized in the design of magnets which are used to produce the homogeneous magnetic field B.sub.o. The use of this geometry imposes two constraints on the design of RF coils to be used in an NMR imaging system. One of these constraints is that the RF coil should be constructed on the surface of a cylinder so there is free access along the axis of the whole to accommodate the patient ILLIGIBLE constraint is that the radio-frequency field B.sub.1 produced by the RF coil must be perpendicular to the solenoidal axis of symmetry which is parallel to the axis of field B.sub.0 (typically selected to be in the Z direction of the Cartesian coordinate system).

Detailed Description Text (3):
FIGS. 1A and 1B depict schematically one conventional NMR coil design. The coil is made up of single turns 1 and 3 connected in parallel and driven at points 7 and 9 across a tuning capacitor 8. Such a coil is typically formed from copper tubing 5 which is mounted on a non-conductive (high dielectric) cylindrical form 11, as seen in FIG. 1B. Each of the coil turns is sized to cover 120.degree. of the cylinder's circumference. The coil region where connections 7 and 9 are made is sized to cover approximately 60.degree. of the circumference. For maximum RF field uniformity, the side of the coils parallel to the longitudinal axis of the cylinder should be equal to two cylinder diameters (D). However, a coil having a side length of two diameters is impractical, because RF energy is placed in regions of the patient which are not of interest. Therefore, in practice, the coil side length is reduced to approximately one diameter length.

Detailed Description Text (4):
FIG. 1C illustrates another embodiment of a conventional RF coil which is similar to that depicted in FIG. 1A, but in which coil turns 15 and 17 are connected in series and driven at points 19 and 21 across a capacitor 18. The coil illustrated in FIG. 1C is typically utilized in NMR studies of the head.

Detailed Description Text (5):
FIG. 1D depicts as yet another embodiment of a conventional NMR RF coil made up of two conductive loop elements 2 and 4 fabricated from copper foil. The loop elements are electrically interconnected by a conductive strip 6. A second conductive strip 8 disposed opposite ILLIGIBLE p6 is electrically connected to one of the loops, for example 2, but is separated at its other end by an air space formed between it and loop 6. The coil is energized across the airgap at points 10 and 12. Current flow is as indicated by arrows 14.

Detailed Description Text (6):
RF coils having a single turn or two turns, as described hereinabove, have been used to minimize the inductance and increase the resonant frequency to permit NMR studies to be performed at higher magnetic field strengths. However, as described hereinbefore, the concentration of the resonant current in so few turns reduces the homogeneity of the B.sub.1 field and homogeneity of the signal sensitivity in the sample volume undergoing examination. Moreover, the lack of symmetry between the position of the timing capacitor and the stray capacitance of, for example, the single-turn coil leads to a non-uniform current distribution in the coil, and a corresponding reduction in the uniformity of the B.sub.1 field. One of the effects of stray capacitance in the low-turn coils (as well as in others) is to cause currents not to circulate through complete coil loops, but to leak through the object undergoing examination. This has the deleterious effect of destroying field uniformity. It will be recognized that increasing the number of series coil turns in an effort to increase homogeneity is not a viable solution, since increased coil turns increase inductance (hence, placing a ceiling on the resonant frequency of the coil).

Detailed Description Text (7):
It is, therefore, apparent that current distribution needs to be controlled in a number of coil windings to produce a uniform B.sub.1 field. Additionally, as already indicated hereinbefore, coil geometry should be such that there is free access along its longitudinal axis for positioning a patient. The B.sub.1 field must also be perpendicular to the cylindrical axis of symmetry which is selected to be parallel to the direction of the B.sub.0 field. The single-turn coil shown in FIG. 2A satisfies these constraints and is the basic element of the coil in accordance with the invention.

Detailed Description Text (8):
Referring now to FIG. 2A, the single turn coil is comprised of two parallel

conductive segments 21a and 22a each having a capacitor 23a connected in series therewith. The ends of conductors 21a and 22a are connected to diametrically opposed points on a pair of parallel conductive loops 25a and 26a spaced apart along common longitudinal axis 16. The coil could be driven by a source such as an RF amplifier generally designated 20 connected between terminals 27a and 28a in parallel with the capacitor in segment 21a. Arrows 29 indicate the relevant current paths which produce a B.sub.1 radio-frequency field perpendicular to the plane defined by conductive wire segments 21a and 22a which, for convenience, will be hereinafter referred to as being vertical. It should be noted that the direction of the B.sub.1 field may be determined by the conventional right-hand rule. The rule states that, if the fingers of the right hand are placed around the current-carrying segment so the thumb points in the direction of current flow, the fingers will point in the direction of the magnetic field (i.e., B.sub.1).

Detailed Description Text (9):
The NMR coil design in accordance with the invention is comprised in the preferred embodiment of a plurality of vertical wire segments 21b evenly spaced and connected around the upper and lower conductive circular loops 25b and 26b as shown in FIG. 2B. It will be recognized that the loops need not be precisely circular but may also be elipsoidal or of some other geometrical form generally having an opening therein to accommodate the object to be examined. Each of the vertical conductive segments includes at least one capacitive element 23b. The multiple current paths, each equivalent to that in FIG. 2A, are indicated by arrows 29 in FIG. 2B, and will be discussed in greater detail hereinafter. The homogeneity of the B.sub.1 field increases as the number of vertical conductive segments is increased. This is due to the fact that, as the number of segments is increased, the resultant field is produced by many contributions so that the effect of any one conductor is reduced. The number of conductors cannot be increased without limit since the open spaces between adjacent vertical conductors are needed to allow a path for the magnetic flux, due to current flow, to escape thereby producing a homogeneous B.sub.1 field. Coils having 4, 8, 16, and 32 vertical conductors have been constructed. It should be noted that the vertical conductive segments need not be evenly spaced. In fact, an embodiment of the inventive RF coil having a window formed therein to facilitate observation of the patient is disclosed hereinafter. What is needed to produce a homogeneous B.sub.1 field is a plurality of vertical conductors distributed around the periphery of the conductive loops such that the current in the vertical conductors approximates a sinusoidal distribution. The resulting inventive NMR coil may be thought of as a resonant cavity made up of an open-ended cylinder with an oscillating magnetic field transverse to the cylinder's axis when the coil is excited by a sinusoidal voltage or current source. There are a number of resonant modes possible, as will be more fully described hereinafter.

Detailed Description Text (10):
A better understanding of the inventive coil depicted schematically in FIG. 2B can be acquired by study of the lumped-element-equivalent circuit for this coil configuration as shown in FIG. 3A. The equivalent circuit is a balanced-ladder network made up of a repeat circuit unit depicted in FIG. 3B and generally designated 30. Each unit is comprised of inductive elements 31 and 32, each having connected at one of the ends thereof a series-connected combination of inductive and capacitive elements 33 and 34, respectively. The two points labelled A (FIG. 3A) are joined together to complete the upper conductive loop 26b, and the points labelled B are joined to complete the lower conductive loop 25b. Inductors 31 and 32 represent the inductance associated with each loop segment 24 along the upper and lower conductive loops of the coil. These inductors are all mutually inductively coupled. Likewise, inductors 33 associated with vertical wire segments 21b are all mutually inductively coupled. To some extent, the vertical wire segments appear to be wired in parallel with a resulting reduced net inductance compared to the single-turn version shown in FIG. 2A. On the other hand, the mutual coupling increases the net inductance of the upper and lower conducting loops (compared to the sum of the individual self inductances in the loops). In practice, the loop and vertical segment inductances can be minimized by constructing both from a wide band of conducting foil. This may be advantageously accomplished by etching the conductors using a flexible printed circuit board. It may be desirable, for example, to minimize inductances 31, 32, and 33 (FIG. 3A) to raise the operating frequency of the coil.

Detailed Description Text (11):
Referring now to FIG. 3B, in operation, the voltage between points E and F is phase shifted with respect to the voltage between points C and D. At the frequency where

the cumulative phase shift for all units 30 (FIG. 3A) adds to 2 .pi. radians, the coil has a standing wave resonance. At this resonance, referred to as the primary resonance, the current in each vertical wire segment 21b has been found to be approximately proportional to sin .theta., where .theta. (see FIG. 2B) is the polar angle of the vertical wire segment measured from the Y axis, for example. Such sinusoidal current distribution produces an increasingly more homogenous transverse field as the number of vertical wire segments 21b is increased.

Detailed Description Text (12):
The coil configuration represented by the lumped-element equivalent circuit of FIG. 3a can also exhibit higher frequency resonances which produce higher order transverse field distributions. The higher resonance frequencies are excited by using an appropriately higher frequency excitation source. These resonances will be referred to as secondary resonances. For example, when the cumulative phase shift around the network equals 4 .pi. radians, the current in the vertical wire segments is proportional to sin 2 .pi.. For this resonance, the X and Y components of the transverse field show an approximately linear gradient along the X and Y axes, respectively, with nulls at the center of the coil.

Detailed Description Text (13):
It is not known whether the particular set of equations describing the lumped-element-equivalent network circuit of FIG. 3a has been solved analytically. However, wave propagation in periodic structures has been studied extensively, particularly in solid-state physics, and provides support for the intuitive description of the current distributions in the lumped-element-equivalent circuit. Connecting the ends (A and B, FIG. 3a) of the ladder to produce upper and lower loops imposes periodic boundary conditions which are also often used in crystal lattice theory. For 2N repeat elements 30, there are 2N+1 loop currents and 2N+10linear equations. One loop current can be set equal to zero provided the current in loops 25b and 26b are balanced. The 2N remaining equations can be represented by a 2N.times.2N Toeplitz matrix which has N pairs of eigensolutions. The eigen currents are proportional to sin n .theta. and cos n .theta. with 1.ltoreq.n.ltoreq.N. Numerical solutions of the equations indicate that the currents are sinusoidal.

Detailed Description Text (14):
Several advantages are realized if the coil is constructed to have four-fold cylindrical symmetry. As used herein, four-fold cylindrical symmetry means that coil geometry (i.e., the position of vertical segments along loop periphery and the capacitive values in each segment) remains the same when the coil is rotated by 90.degree. about its longitudinal axis. For example, coils having a number of vertical segments which is a multiple of four (e.g., 4, 8, 12, 16, 32) have four-fold symmetry. In this case, the primary resonance has two orthogonal degenerate modes at the same resonant frequency. One mode, referred to herein as the X mode, gives an RF magnetic field parallel to the X axis when the current in the vertical wires is proportional to sin .theta.. For the other mode, referred to as the Y mode, current is proportional to cos .theta. and the magnetic field is in the Y-axis direction. If the RF coil is driven by applying power from an RF amplifier (not shown) at a single point, such as between terminals 27b and 28b depicted in FIG. 2b, only the X mode is excited. The resonant circuit in this case produces an oscillating RF field 2H.sub.1 cos .omega.t, which can be thought of as two fields, each H.sub.1 in magnitude, rotating in opposite directions in the transverse plane perpendicular to the direction of the B.sub.o field. The nuclei respond to only one of the two rotating fields. Specifically, the nuclei respond to the field rotating in the direction in which the polarized nuclei precess. Hence, the power used to create the B.sub.1 component rotating in the wrong direction is wasted. However, if as shown in FIG. 4, the coil is powered at a second drive point in vertical conductor 41 located 90.degree. from the first driving point in vertical conductor 40 with a source 90.degree. out of phase, the two oscillating fields add vectorially to give a single rotating field. In this case, no driving power would be wasted. Hence, driving the inventive RF coil at two driving points separated by 90.degree. doubles the RF power efficiency. Also, because the noise voltage generated in the two orthogonal modes are not correlated but the signals from the nuclei are correlated, the signal-to-noise ratio can be enhanced by a factor of the .sqroot.2. In this case, the detected NMR signal must be sampled at the two orthogonal points of the coil.

Detailed Description Text (15):
The need to maintain orthogonality for the two degenerate X- and Y-resonant modes

places constraints on component tolerances and coil geometry. For example, the effective coefficient of inductive coupling, K, between the two modes must be kept small compared to the reciprocal of the coil, quality factor Q. At high frequencies, where patient loading of the coil is high and increased RF power eficiency is more desirable, a lower Q of the coil relaxes somewhat the constrains on achieving orthogonality. The two resonance modes are substantially uncoupled if the product K.times.Q is less than about 5%. In this case, each mode will have the correct phase shift to produce a rotating field.

Detailed Description Text (16):
The directions of currents in vertical and loop conductors for a coil having eight vertical conductors, and hence four-fold symmetry, are indicated by arrows 29 in FIG. 2B which depicts such a coil. The current directions are for the primary (desired) resonance mode. The sinusoidal nature of these currents will now be discussed in greater detail with reference to FIGS. 5A-5C. Referring now to FIG. 5A, there is shown a top view of the coil depicted in FIG. 2B. The coil is energized at points 27b and 28b, as before, which are in a segment arbitrarily assigned to a position .theta.=0.degree.. With the coil energized in this manner, the maximum current proportional to cos .theta. flows in the segment located at .theta.=0.degree. in a direction out of the paper plane as suggested by the circled dot: Smaller currents (proportional to cos .theta., wherein .theta.=45.degree. and 315.degree.) flow in the same direction in the segments adjacent the one situated at .theta.=0.degree.. Currents of corresponding magnitude flow in an opposite direction (into the paper, as indicated by the circled cross) in the segments situated at .theta.=180.degree., 135.degree., and 225.degree.. The magnitude of current flow in the conductive segments is graphically depicted in FIG. 5B, in which position angle .theta. is indicated along the horizontal axis, while current magnitude is indicated along the vertical axis. Currents flowing out of the paper (45.degree., 0.degree., 315.degree.) have been arbitrarily designated as having positive values, while those flowing into the paper (135.degree., 180.degree., 225.degree.) have negative values. In the primary resonant mode, the segments at .theta.=90.degree. and .theta.=270.degree. do not conduct any current and in practice may be eliminated or replaced by short circuits.

Detailed Description Text (17):
The direction of current flow in upper conductive loop 26b (FIG. 5A) is indicated by arrows 50 which are sized relative to one another to indicate approximate magnitudes. More procisely, loop current distribution is graphically depicted in FIG. 5C with angular position and current magnitude being indicated along horizontal and vertical axes, respectively, and wherein clockwise current flow is arbitrarily assigned to have a positive value. The loop currents are distributed in a step-wise manner. Thus, currents flowing between 45.degree. and 90.degree., and between 315.degree. and 270.degree. are larger than those between 0.degree. and 45.degree., and between 0.degree. and 315.degree., respectively, since the former include currents provided by segments at 45.degree. and 315.degree..

Detailed Description Text (18):
In some coil embodiments (see FIG. 5D), particularly those used for NMR imaging of the head, it has been found advantageous to cut a window in the coil form to provide ready means for the patient face to be visible. This has necessitated the removal of some of the vertical segments to provide an unobstructed area in which to cut the window. This is especially true of coils having closely spaced vertical segments, such as the 32 segments in an embodiment disclosed hereinafter. To minimize the perturbation of RF field homogeneity, it has been found desirable to remove those segments which carry little or no current. In the embodiment shown in FIG. 5A, either one of the conductors situated at 90.degree. or 270.degree. could be removed without significantly affecting field homogeneity. The current distribution in the vertical segments in the case where the segment at 90.degree. is removed is graphically in FIG. 5E. The current magnitude at points greater than 45.degree. and less than 135.degree. is zero.

Detailed Description Text (19):
In the embodiment having 32 vertical segments, six segments were removed to accommodate the window. The coil was found to work satisfactorily without any adjustment. In the preferred embodiment, however, it has proven advantageous to increase the capacitance values in the segments nearest the window on either side to accommodate increased currents therein to compensate for the eliminated current-carrying capacity of the removed segments.

Detailed Description Text (20):
In some head coil embodiments, it may also be desirable to bend one of the loop elements into a generally saddle-shaped configuration. The raised loop sections (J and K in FIG. 5D) fit over the shoulders allowing the head to be more fully enclosed by the coil.

Detailed Description Text (21):
It has also been found advantageous in the case of NMR head coils to fabricate the inventive coil on two separable coil assemblies as claimed and disclosed in commonly assigned co-pending application Ser. No. (15-NM-2442), which is incorporated herein by reference as background material. In this case, it is necessary to form open circuits in the upper and lower loop conductors at points X and Z as indicated in FIG. 6A. The coil is still energized at a point .theta.=0, such that in operation the two coil halves resulting from the open circuits are coupled by mutual inductance to operate as a single coil. FIG. 6A is similar to FIG. 5A, with the exceptions that, due to the loop open circuits, segments at 90.degree. and 270.degree. carry oppositely directed currents. The segment current distribution for this embodiment is depicted in FIG. 6B, and, as before, has a sinusoidal geometry dependent on angle .theta. with maximum current occuring in segments near .theta.=0.degree., 45.degree., 180.degree., and .theta.=225.degree., FIG. 6C depicts the current distribution in the conductive loop elements. Maximum loop current values occur at values of 0 slightly greater than 90.degree. and 270.degree. at points designated W.

Detailed Description Text (22):
If desired, a window may be formed in the coil embodiment depicted in FIG. 6A, by removing conductive segments carrying the lowest currents. Such segments are located between 90.degree. and 135.degree., and 270.degree. and 315.degree. as indicated by regions designated W in FIG. 6A. Reference to FIG. 6B will indicate that these regions correspond to segments having the lowest currents and, therefore, would have the least impact on RF field homogeneity.

Detailed Description Text (23):
If the coil geometry (i.e., either the location of vertical segments along the loop periphery or the capacitance values of the capacitors in each segment) is selected to have other than four-fold symmetry, the X- and Y-resonant modes are orthogonal and occur at different frequencies. One method of exciting the two resonances is with two sources, as described hereinbefore. It is, however, possible to excite two resonances using a single source having the needed frequency components. Small variations in component value or coil geometry may give two overlapping resonances if the coil Q is high enough. This may be troublesome if only a single resonance is desired. One of the two resonances may be sufficiently displaced, however, in frequency to render it harmless if the coil symmetry is grossly perturbed. One possibility is to cut the upper and lower conductive rings 25b and 26b (FIG. 2b) at the points where the desired mode has current nulls. Another possibility is to replace those capacitors 23b which carry no current in the desired mode by short or open circuits. A short-circuited segment in effect appears as a large capacitance and therefore tends to lower the resonant frequency of the undesired mode. The effect of an open circuit is to decrease the apparent capacitance to thereby increase the resonant frequency.

Detailed Description Text (24):
It may be advantageous in the non-four-fold symmetrical coil to manipulate the two resonant frequencies by proper choice of capacitors 23b. The X mode has maximum currents where the Y mode has minimum current. Hence, by increasing the capacitor values where sin 0 is large and decreasing the capacitors where cos .theta. is large, the X-mode frequency can be lowered and the Y-mode frequency raised. Such a coil would be useful for performing simultaneous NMR double resonance studies. For example, one mode could be timed for the proton .sup.1 H resonance and the other mode for the fluorine .sup.19 F resonance.

Detailed Description Text (25):
There are a number of ways that the inventive NMR coil design in principle can be implemented for in vivo NMR studies. In the preferred embodiment, the conductive elements (e.g., 21b, 25b and 26b, FIG. 2b) are constructed with wide sheets of conductive foil in order to minimize their self inductance. They also could be constructed with large diameter conductive tubing, for example. The distance between the upper and lower conductive rings should be about one or more times the coil diameter to reduce field inhomogeneity due to the currents in loops 25b and 26b.

Detailed Description Text (26):
If a coil is required to resonate at a single predetermined frequency, it is possible to construct a coil patterned after FIG. 2b using only fixed capacitors. It is, however, more practical to include some variable elements for fine tuning the resonant frequency. The minimum requirement for tuning both X and Y modes is to place a variable trimming capacitor in each of two vertical conductive elements located 90.degree. apart (e.g., 40 and 41, FIG. 4). Small perturbations on the capacitance at these two points will not greatly disturb the field homogeneity.

Detailed Description Text (27):
Where a wider adjustment of resonant frequency is desired, it is preferable to tune all of the capacitors simultaneously or to change the effective inductance of the coil assembly. Small variations in inductance can be achieved by varying the width of foil-conductive elements. Larger variations in inductance can be achieved by varying the lengths of the vertical conductors by adjusting the distance between the two conductive loops.

Detailed Description Text (28):
The manner in which the preferred embodiment of an inventive coil having 32 segments and which was physically and electrically sized for NMR head studies will now be described with reference to FIGS. 7A and 7B. The same construction method is utilized in the construction of body coils which are typically sized to have a larger diameter. The head coil was operable at a frequency of 21.31 MHz, which is determined by the strength of main field B.sub.o, and the NMR isotope studied. In general, the coil is fabricated by etching (using conventional techniques) four double-sided copper-clad Teflon resin printed circuit boards. The boards are mounted on a cylindrical form having a 10.5 inch outside diameter. Each side of the circuit boards is etched with a different conductive pattern. Each circuit board is approximately 8 by 12 inches.

Detailed Description Text (31):
The inner and outer etched surfaces are overlayed such that points S, T, U, V (FIG. 7A) lie above points O, P, Q, R (FIG. 7B), respectively. In this manner, gaps 77 on each etched (inner and outer) surface are bridged by continous portions of the unconnected two-thirds 83 of the straight elements 75 on each surface. Gaps 79 are bridged by continous portions 81 of the straight element. The combination of cooper foil segments and printed circuit dielectric form three series connected capacitors along the length of each straight conductor. The number of capacitors can be varied by increasing or decreasing the number of gaps. The net capacitance in each straight conductor is typically adjusted to be approximately equal. The adjustment is accomplished by electrically connecting one or more of copper pads 85 to change the area of the overlap of the inner and outer surfaces. In the preferred embodiment, the inner and outer patterns are etched on opposite sides of a double-sided printed circuit board.

Detailed Description Text (32):
The inner and outer etched surfaces of strips 71 and 73 are electrically connected together at points O and S, P and T, Q and U, and R and V. A complete coil requires four such overlayed and interconnected assemblies. A half of the coil is made by electrically joining two assemblies. Points O and Q of one quarter assembly are electrically connected to points P and R, respectively, of the second quarter assembly. The two coil halves constructed in this manner are mounted on a cylindrical coil form without electrical connections between them. Leaving the two halves of the loop conductors disconnected splits the degencracy of the two desired resonances, as disclosed hereinbefore. The two coil halves are coupled, in operation, by the mutual inductances thereof when one of the halves is energized across one of the three capacitors in a straight conductor, such as, for example, at points 89 and 91 shown in FIG. 7A. The drive point impedance was about 50 ohms without any adjustment with a patient's head positioned in the coil (i.e., with a loaded coil).

Detailed Description Text (33):
In the preferred embodiment, the double-sided printed circuit board dielectric (Teflon resin) thickness was about 0.006". Each of the three capacitors in each straight conductor was adjusted to equal approximately 133 pico farads. It should be noted that it is not important that each capacitor have equal value, but only that the net capacitance of each straight conductor are equal. The desired resonance frequency with a homogeneous RF magnetic field was at 21.31 MHz.

Detailed Description Text (34):
Another embodiment of an inventive NMR coil was constructed following the patterns disclosed with reference to FIGS. 7A and 7B and having 32 vertical segments. This coil was constructed on a cylindrical form having an outside diameter of 11.5 inches and a length of 16.5 inches. Strip elements 71 and 73 (FIGS. 7A and 7B) were 0.25 of an inch wide. Straight conductors 75 were 0.5 of an inch wide spaced at five-eights inch intervals. In this case, there were ten gaps in each straight conductor, similar to gaps 77 and 79, so that the value of each capacitor was lower than that in the embodiment of FIGS. 7A and 7B. The coil resonant frequency was 63.86 MHz.

Detailed Description Text (36):
FIG. 8A depicts one coil embodiment comprised of a plurality of vertical conductors 101 equally spaced around the periphery of interconnecting conductive loops 102 and 103. Each of conductors 101 is provided with a variable capacitor 104 built into the lengths thereof. FIG. 8B shows a detailed longitudinal section of one variable capacitor illustrating one possible construction. The capacitor is formed of an inner conductor portion 101a, one end of which extends into a hollow portion 107 of conductor section 101b. The other ends of conductors 101a and 101b are connected to conductive loops 102 and 103. Portion 101a is separated from the inner surface of the hollow section of conductor 101b by means of a sleeve formed of a dielectric material 108 which may comprise quartz or another suitable dielectric material, such as Teflon synthetic resin polymer. A typical number of vertical conductors 101 and, hence, capacitors 104, may be selected (but is not limited to) to be between 8 and 32. All of the capacitors can be tuned simultaneously by changing the length (or height) of the device. The change in inductance with length is a smaller effect than the change in capacitance. It will be recognized, of course, that each vertical conductor (in either this embodiment or in those described below) need not have a variable capacitance associated therewith, if only a single resonant frequency is desired.

Detailed Description Text (37):
FIGS. 9A and 9B depict another embodiment of the inventive NMR RF coil which is tuned by varying the inductance rather than capacitance. The coil is comprised of a plurality of parallel vertical conductors 110, each having a pair of fixed capacitors at each of the ends thereof. The conductors are evenly spaced about the circumference of a pair of parallel conductive loops 112 and 113 but are electrically insulated therefrom to form a pair of capacitors at the ends. FIG. 9B illustrates in detail the manner in which the capacitors are formed. Each of conductors 110 extends through openings 114 formed in a loop 112, for example. Conductors 110 are electrically insulated from the conductive loop by sleeves 115 made of a dielectric material lining openings 114. The coil is simultaneously tuned by moving one or both of end loops 112 and 113 closer or farther apart. This changes the inductance without changing the capacitance of the coil.

Detailed Description Text (38):
Another embodiment of the inventive NMR RF coil is schematically depicted in FIG. 10. In this embodiment, a plurality of parallel conductors 120, electrically shorted at one of the ends 121 thereof, are plated (or etched) on the outside surface of a dielectric cylindrical form 122. A similar cylindrical form 124 having a slightly smaller diameter than that of form 122 also has plated thereon a plurality of parallel conductors 125 shorted at their ends 126. Cylindrical form 124 having a smaller diameter than that of form 122 is adapted to be slidably inserted thereinto, such that the unshorted ends of conductors 120 and 125 overlap. The capacitance between corresponding matching wires depends on degree of overlap between conductors 120 and 125. The device is tuned either by sliding one form in or out when the wires are aligned (varying the inductance and capacitance) or by rotating one form slightly with respect to the other to misalign the conductors to thereby vary the capacitance.

Detailed Description Text (39):
FIG. 11 depicts as yet another embodiment of the inventive NMR RF coil in which parallel isolated conductors 130 are plated onto the outside surface of a dieloctric cylindrical form 132. A pair of conductive loops 133, 134 are then inserted into each of the ends of the cylindrical form to a position designated by the dotted lines 135 and 136, such that each loop couples capacitively to the wires through the dielectric material comprising the cylinder walls. Moving the loops in or out along the longitudinal axis of the cylinder tunes the device by varying the degree of overlap between the loop and the conductors and thereby varies the capacitance. The

device may also be tuned inductively by sliding at least one of the loops so as to vary the length of vertical conductors 130. It will be recognized that conductors 130 could be positioned on the inside of the cylindrical form while the conductive loops would be inserted onto the outside of the cylinder.

Detailed Description Text (40):
In each of the above-described exemplary NMR coil embodiments, the tuning is accomplished by varying the degree of overlap or relative length of the conductive coil elements. The relative motion necessary to vary the degree of coupling or the conductor length occurs at capacitive coupling points. Hence, there is no contact resistance to create noise or losses.

Detailed Description Text (41):
It will also be recognized that it is desirable for efficient power transfer to have matching coil input and transmitter output inpedances. In the inventive NMR coil configuration, this may be accomplished by providing a plurality of series-connected capacitances, such as capacitances 150-154, in a vertical conductor 148, as shown in FIG. 12. In this case, an appropriate pair of terminals 155-160 providing the desired impedance is selected as needed to provide the best match to the transmitter impedance. When two driving points are used to energize the coil, similar series capacitances 161-165 may be utilized in the second vertical conductor 149 which is perpendicular to the first driven conductor. In this case, variable capacitances 166 and 167 in conductors 148 and 149, respectively, are utilized to fine tune the coil. In those vertical conductors which are not used as driving points, the capacitance needed to resonate the circuit need not be distributed in a string of series-connected capacitors, but may instead be lumped into a single capacitance 168 shown in FIG. 12 as being connected in a vertical conductor 147, which may also include a variable tuning capacitor 169. As in the case of FIG. 4, some of the vertical conductors in FIG. 12 are not shown to preserve clarity.

Detailed Description Text (42):
From the foregoing, it will be appreciated that, in accordance with the invention, an NMR RF coil is provided in which the current and tuning capacitance are distributed in many turns, but in which the effective inductance is approximately equal to or less than that of a single-turn coil. The inventive NMR RF coil also provides a considerable improvement in the uniformity of the B.sub.1 field and in signal sensitivity. The coil's geometry also permits improvements in signal-to-noise ratio and reduction in RF driving power.

Other Reference Publication (1):
Journal of Magnetic Resonance 35, No. 3, pp. 329-336 (Sep. 1979), "Simultaneous Multinuclear NMR by Alternate Scan Recording of .sup.31 P and C Spectra"; Styles.

Other Reference Publication (2):
Nuclear Magnetic Resonance Imaging in Medicine--Igaku-Shoin, 1981, Leon Kaufman, pp. 62-64.

CLAIMS:

1. An NMR radio-frequency (RF) coil comprising:

a pair of conductive loop elements disposed in a spaced apart relation along a common longitudinal axis; and

a plurality of conductive segments each having at least one reactive element in series therewith, said segments electrically interconnecting said loop elements at points spaced along the periphery of each of said loops, said segments being disposed substantially parallel to said longitudinal axis such that the resulting configuration has four-fold symmetry, wherein in operation said RF coil is capable of producing an RF field useful for performing NMR studies.

2. The NMR RF coil of claim 1 wherein said conductive segments are spaced at equal intervals along the periphery of each of said loops.

3. The NMR RF coil of claim 1 or 2 wherein said reactive element comprises at least one capacitive element.

4. The NMR RF coil of claim 3 wherein said coil is energizable across at least one of said capacitive elements in one of said conductive segments to achieve in

operation a current in said segments approximating a sinusoidal distribution dependent on the segment angular position .theta. along the loop peripheries.

5. The NMR RF coil of claim 3 wherein said coil is energizable by a first source connected in parallel with one of said capacitive elements in a first one of said segments so as to excite said RF coil to produce an oscillating RF field perpendicular to said longitudinal axis and described by 2H.sub.1 cos .omega.t, in which

H.sub.1 is the magnitude of the oscillating RF field,

.omega. is the resonant frequency of the RF field,

t is time,

said RF field being made up of two component rotating in opposite directions.

6. The NMR RF coil of claim 5 wherein said coil is energizable by a second source 90.degree. out of phase with said first source, said second source being connected in parallel with a capacitive element in a second one of said segments separated by an angle .theta.=90.degree. from said first segment such that said orthogonal RF field components add vectorially to produce a single rotating RF field wherein .theta. is a polar angle indicative of the position of each of said segments along the periphery of said loops.

7. The NMR RF coil of claim 3 wherein each of said segments comprises first and second sections, said sections being connected at one end to a respective one of said loop elements, the other end of said first sections having formed therein a recess for slidably receiving the other end of a corresponding one of said second sections, said other ends of said first and second sections being separated by a dielectric material, said segments being moveable relative to one another thereby to form a variable capacitive element.

8. The NMR RF coil of claim 3 comprising a first and second hollow cylindrical form of dielectric material, said forms being sized for telescoping movement relative to one another, said segments being composed of first and second sections supported by a major surface of said first and second forms, respectively, said first and second conductor sections being connected at one of the ends thereof to a respective one of said loop elements, the other ends of matching ones of first and second sections being positioned to overlap one another so as to form a capacitive element, the degree of overlap therebetween being adjustable by relative movement of said forms.

9. The NMR RF coil of claim 3 comprising:

a hollow cylindrical form of dielectric material, said segments being supported by a major surface thereof parallel to said longitudinal axis; and

said pair of conductive elements being disposed on a surface opposite to the surface of said form supporting said segments to form a capacitive element in the region of overlap between said loops and said segments, at least one of said loop elements being slidably disposed on said form so as to vary the region of overlap thereby to vary the capacitance of said capacitive element.

10. The NMR RF coil of claim 3 wherein at least one of said conductive loop elements includes a plurality of spaced apertures formed around the periphery thereof for slidably receiving corresponding ones of said segments, said apertures being lined with a dielectric material to electrically insulate said segments from said loop element so as to form fixed-value capacitive elements, the inductance of the coil being adjustable by relative motion of said loop element having apertures formed therein along said segments to vary the length thereof.

11. The NMR RF coil of claim 3 wherein of said segments each includes at least one variable capacitive element.

12. An NMR radio-frequency (RF) coil comprising;

a pair of conductive loops disposed in a spaced-apart relation along a common longitudinal axis; and

a plurality of conductive segments each having at least one capacitive element in series therewith, said segments electrically interconnecting said loop elements at points spaced along the periphery of each of said loops, said segments being disposed substantially parallel to said longitudinal axis resulting in an NMR RF coil having non-four-fold symmetry wherein in operation said RF coil is capable of producing an RF field useful for performing NMR studies.

13. The NMR RF coil of claim 12 wherein said conductive segments are spaced at equal intervals along the periphery of each of said loops.

14. The NMR RF coil of claim 12 wherein said coil comprises an NMR RF head coil and wherein one of said conductive loop elements has a geometry fitable about the shoulders of a subject allowing the head to be more fully positioned within the coil.

15. The NMR RF coil of claim 12 including means allowing said coil to be energized across at least one of said capacitive elements in one of said conductive segments to achieve in operation a current in said segments approximating a sinusoidal distribution dependent on the segment angular position .theta. along the loop peripheries.

16. The NMR RF coil of claim 15 wherein said means includes first and second means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes having segment current distribution proportional to sin .theta. and cos .theta., respectively, said capacitive elements in segments where sin .theta. is large having high capacitive values, while said capacitive elements in segments where cos .theta. is large having lower capacitive values relative to said high values such that said first and second modes occur at different frequencies.

17. The NMR RF coil of claim 15 including means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes, one of said modes being a desired mode, said pair of conductive loop elements each including an open circuit along the peripheries thereof where the desired mode has current minima so as to displace the frequency of the undesired mode relative to the frequency of the desired mode.

18. The NMR RF coil of claim 15 including means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes, one of said modes being a desired mode, said capacitive elements in segments carrying negligible current in the desired mode being replaced by short circuits so as to displace the frequency of the undesired mode relative to the frequency of the desired mode.

19. The NMR RF coil of claim 15 including means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes, one of said modes being a desired mode, said capacitive element in segments carrying negligible current in the desired mode being replaced by open circuits so as to displace the frequency of the undesired mode relative to the frequency of the desired mode.

20. The NMR RF coil of claim 19 wherein said coil comprises a head coil and wherein said conductive loop and segments are mounted on a substantially cylindrical coil form, which form is provided with a window in the region where said segments have been replaced by open circuits.

21. The NMR RF coil of claim 15 wherein said means comprises first and second means disposed on first and second segments, respectively, separated by an angle .theta.=90.degree..

22. The NMR RF coil of claim 26 wherein said first and second means each comprise a plurality of series-connected capacitive elements, the common points between said series-connected capacitive elements being selectable to adjust the input impedance of said RF coil.

23. The NMR RF coil of claim 15 wherein said means comprises a plurality of series-connected capacitive elements, the common points between said series-connected capacitive elements being selectable to adjust the input impedance of said RF coil.

24. The NMR RF coil of claim 13 wherein each of said segments comprises first and second sections, said sections being connected at one end to a respective one of said loop elements, the other end of said first sections having formed therein a recess for slidably receiving the other end of a corresponding one of said second sections, said other ends of said first and second sections being separated by a dielectric material, said segments being moveable relative to one another thereby to form a variable capacitive element.

25. The NMR RF coil of claim 13 comprising a first and second hollow cylindrical form of dielectric material, said forms being sized for telescoping movement relative to one another, said segments being composed of first and second sections supported by a major surface of said first and second forms, respectively, said first and second conductor sections being connected at one of the ends thereof to a respective one of said loop elements, the other ends of matching ones of first and second sections being positioned to overlap one another so as to form a capacitive element, the degree of overlap therebetween being adjustable by relative movement of said forms.

26. The NMR RF coil of claim 13 comprising:

a hollow cylindrical form of dielectric material, said segments being supported by a major surface thereof parallel to said longitudinal axis; and

said pair of conductive elements being disposed on a surface opposite to the surface of said form supporting said segments to form a capacitive element in the region of overlap between said loops and said segments, at least one of said loop elements being slidably disposed on said form so as to vary the region of overlap thereby to vary the capacitance of said capacitive element.

27. The NMR RF coil of claim 13 wherein at least one of said conductive loop elements includes a plurality of spaced apertures formed around the periphery thereof of slidably receiving corresponding ones of sail segments, said apertures being lined with a dielectric material toelectrically insulate said segment from said loop element so as to form fixed-value capacitive elements, the indoctance of the coil being adjustable by relative motion of said loop element having apertures formed therein along said segments to vary the length thereof.

28. The NMR RF coil of claim 13 wherein at least one of said segments includes at least one variable capacitive element.

29. An NMR radio-frequency (RF) coil comprising:

a pair of conductive loop elements disposed in a spaced-apart relation along a common longitudinal axis; and

a plurality of conductive segments each having at least one reactive element associated therewith, said segments electrically interconnecting said loop elements at points spaced along the periphery of each of said loop elements, said RF coil having means for being energized to achieve in operation, an approximately sinusoidal current distribution in said segments, said distribution being dependent on the angular position .theta. of each of said segments along the loop peripheries.

30. The NMR RF coil of claim 29 wherein said conductive segments are spaced at equal intervals along the periphery of each of said loops.

31. The NMR RF coil of claim 29 wherein said coil comprises an NMR RF head coil and wherein one of said conductive loop elements has a geometry fitable about the shoulders of a subject allowing the head to be more fully positioned within the coil.

32. The NMR RF coil of claim 29 wherin said reactive element comprises at least one capacitive element.

33. The NMR RF coil of claim 32 wherein said means includes first and second means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes having segment current distributions proportional to sin .theta. and cos .theta., respectively, said capacitive elements in segments where

sin .theta. is large having high capacitive values, while said capacitive elements in segments where cos .theta. is large having lower capacitive values relative to said high values such that said first and second modes occur at different frequencies.

34. The NMR RF coil of claim 32 including means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes, one of said modes being a desired mode, said pair of conductive loop elements each including an open circuit along the peripheries thereof where the desired mode has current minima so as to displace the frequency of the undesired mode relative to the frequency of the desired mode.

35. The NMR RF coil of claim 32 including means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes, one of said modes being a desired mode, said capacitive elements in segments carrying negligible current in the desired mode being replaced by short circuits so as to displace the frequency of the undesired mode relative to the frequency of the desired mode.

36. The NMR RF coil of claim 32 including means allowing said RF coil to be energized so as to excite therein first and second orthogonal resonant modes, one of said modes being a desired mode, said capacitive elements in segments carrying negligible current in the desired mode being replaced by open circuits so as to displace the frequency of the undesired mode relative to the frequency of the desired mode.

37. The NMR RF coil of claim 36 wherein said coil comprises a head coil and wherein said conductive loop and segments are mounted on a substantially cylindrical coil form, which form is provided with a window in the region where said segments have been replaced by open circuits.

38. The NMR RF coil of claim 32 wherein said means comprises first and second means disposed on first and second segments, respectively, separated by an angle .theta.=90.degree..

39. The NMR RF coil of claim 38 wherein said first and second means each comprise a plurality of series-connected capacitive elements, the common points between said series-connected capacitive elements being selectable to adjust the input impedance of said RF coil.

40. The NMR RF coil of claim 32 wherein said means comprises a plurality of series-connected capacitive elements, the common points between said series-connected capacitive elements being selectable to adjust the input impedance of said RF coil.

41. An NMR RF coil comprising:

a first assembly having a pair of conductive loop elements disposed in a spaced apart relation along a common longitudinal axis and a plurality of conductive segments electrically connected at the ends thereof to said loop elements, each of said segments having at least one non-conductive gap formed therein; and

a second assembly, substantially identical to said first assembly, in which the gaps formed in the conductive segments are offset relative to the gaps in said first assembly;

said first and second assemblies being disposed coaxially relative to one another and being separated by a dielectric material, the loop elements of one assembly being electrically interconnected to a corresponding loop element of the other assembly, the gaps in each assembly being situated such that a gap in one assembly is bridged by a continuous portion of corresponding segment in the other assembly so as to form a capacitive element.

42. The NMR RF coil of claim 41 wherein the conductive segments in at least one of said first and second assemblies includes at least one electrically insulated conductive pad in the region of the nonconductive gap, said pad being electrically connectable to the remainder of the conductive segment to vary the area of overlap between corresponding segments of said first and second assemblies so as to adjust the capacitance of said capacitive element.

43. The NMR RF coil of claim 41 wherein said first and second assemblies comprise conductive patterns corresponding to said conductive loop and segment elements fabricated on opposite sides of a printed circuit board.

44. The NMR RF coil of claim 41 wherein said first and second assemblies comprise conductive patterns corresponding to said conductive loop and segment elements fabricated on opposite sides of a single printed circuit board.

45. The NMR RF coil of claim 44 wherein said first and second assemblies each comprise a plurality of subassemblies, each subassembly having fabricated thereon a fraction of the conductive pattern associated with one of said assemblies, which subassemblies are electrically interconnected at the conductive pattern portions thereof corresponding to said loop elements to form a complete assembly.

46. The NMR RF coil of claim 44 wherein each of said conductive loop elements includes a pair of open circuits formed therein so as to create two coil halves such that when, in operation, one of said halves is energized the other one is coupled thereto by mutual inductance.

47. The NMR RF coil of claim 41 including means allowing said coil to be energized across at least one of said capacitive elements in one of said conductive segments to achieve in operation a current in said segments approximating a sinusoidal distribution dependent on the segment angular position .theta.0 along the loop peripheries.

48. The NMR RF coil of claim 47 wherein said capacitive elements in segments positioned along the loop peripheries where said sinusoidal current distribution approximates a minimum magnitude comprise a short circuit.

49. The NMR RF coil of claim 47 wherein said capacitive elements in segments positioned along the loop peripheries where said sinusoidal current distribution approximates a minimum magnitude comprise an open circuit.

50. The NMR RF coil of claim 49 wherein said coil comprises a head coil and wherein said conductive loop and segments are mounted on a substantially cylindrical coil form, which form is provided with a window in the region where said segments have been replaced by open circuits.

51. The NMR RF coil of claim 47 wherein said means comprises first and second means disposed on first and second segments, respectively, separated by an angle .theta.=90.degree..

52. The NMR RF coil of claim 51 wherein said first and second means each comprise a plurality of series-connected capacitive elements, the common points between said series-connected capacitive elements being selectable to adjust the input impedance of said RF coil.

53. The NMR RF coil of claim 47 wherein said means comprises a plurality of series-connected capacitive elements, the common points between said series-connected capacitive elements being selectable to adjust the input impedance of said RF coil.

54. The NMR RF coil of claim 41 wherein said coil comprises an NMR RF head coil and wherein one of said conductive loop elements has a geometry fitable about the shoulders of a subject allowing the head to be more fully positioned within the coil.

55. The NMR RF coil of claim 41 wherein said conductive segments are spaced at equal intervals along the periphery of each of said loops.

56. The NMR RF coil of claim 55 wherein said segments are spaced along the periphery of each of said loops to form a configuration having four-fold symmetry.

57. The NMR RF coil of claim 55 wherein said segments are spaced along the periphery of each of said loops to form a configuration not having four-fold symmetry.